ISBN 0-444-81202-4

ANDERSON, HOBART
DANOFF
editors

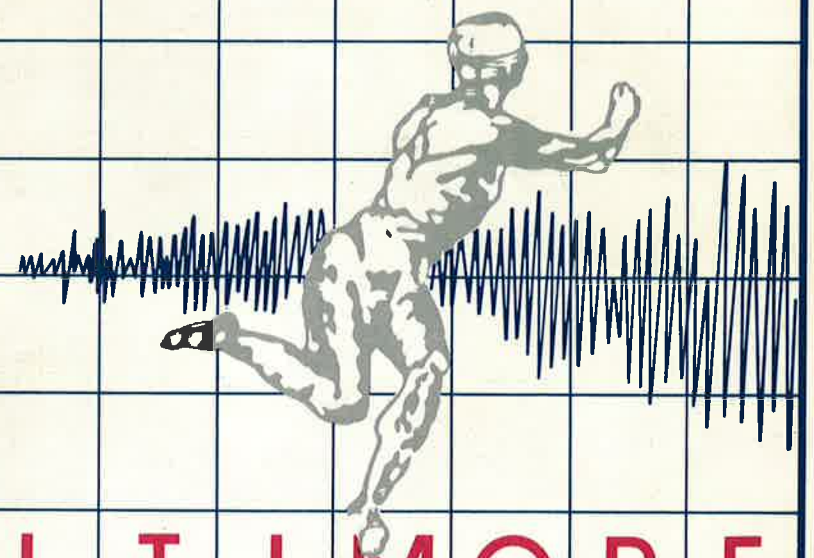
ELECTROMYOGRAPHICAL
KINESIOLOGY

ICS 949

INTERNATIONAL CONGRESS SERIES 949

ELECTRO-MYOGRAPHICAL KINESIOLOGY

editors:
P.A. ANDERSON, D.J. HOBART
J.V. DANOFF



BALTIMORE
1965 - 1990

EXCERPTA MEDICA

ELECTROMYOGRAPHICAL KINESIOLOGY

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Proceedings of the 8th Congress of the International Society of Electrophysiological Kinesiology, held in Baltimore, Maryland, 12–16 August 1990 (25th Anniversary)

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1991

EXCERPTA MEDICA, Amsterdam - New York - Oxford

International Congress Series No. 949
ISBN 0 444 81202 4

This book is printed on acid-free paper.

Published by:
Elsevier Science Publishers B.V.
(Biomedical Division)
P.O. Box 211
1000 AE Amsterdam
The Netherlands

Sole distributors for the USA and Canada:
Elsevier Science Publishing Company Inc.
655 Avenue of the Americas
New York, NY 10010
USA

Printed in The Netherlands

PREFACE

The International Society of Electrophysiological Kinesiology (ISEK) is a multidisciplinary organization composed of members from health related fields such as engineering, physical education, physical therapy and many other disciplines. These clinicians and basic scientists are bound together by a common desire to study human movement and the neuromuscular system.

Every two years, these scientists gather at a Congress of ISEK to share advances and knowledge in the broad field of Electrophysiological Kinesiology. The 8th Congress and 25th Anniversary provided such a forum. The founding president, Dr. John Basmajian, set the tone for the congress in his plenary speech 'The Pivotal Role of ISEK in Biomedical Sciences'. Eleven invited speakers provided a 'state of the art' review in each of the individual scientific sessions. One hundred thirty-one platform papers added to our current knowledge. The following proceedings are a selection of the invited and platform papers.

The participants which affiliated with research institutes and universities bring experience in the fields of Electromyography, Functional Electrical Stimulation (FES), Motor Unit Control, Neuromuscular Diseases, Rehabilitation, Muscle Fatigue, Kinesiology, Motion Analysis and Ergonomics. This book is offered as a contribution to the body of knowledge for the neuromuscular system in normal and diseased states. It includes both basic and applied research.

The status quo is changing fast. It is estimated that man's knowledge is now doubling every five to ten years. Thus, these authors are a crucial part of the information explosion in the field of the Electrophysiological Kinesiology.

Paul A. Anderson
Donald J. Hobart
Jerome V. Danoff

ACKNOWLEDGEMENTS

The papers of this book were presented at the 8th Congress and 25th Anniversary of the International Society of Electrophysiological Kinesiology, August 12-16, 1990, Baltimore, Maryland, USA. The value of this book was established by the many excellent researchers that participated in the Congress. We thank them for their contributions.

Many people have given time and support towards the success of this congress. They are too numerous to mention; however two individuals must be recognized. Without the untiring work of Mary R. di Sabatino and Tammy Marshall, the scientific program, this volume and the Congress would not have become a reality. Additional thanks must go to Jon Laking for this superb work with the Congress audiovisual equipment. We thank especially the members of the program committee for their contribution in refereeing the abstracts submitted for the congress:

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PLENARY OPENING ADDRESS

PIVOTAL ROLE OF ISEK IN BIOMEDICAL SCIENCES

John V. Basmajian

A 25th Anniversary plenary address devoted to the past, present, and future must be all things to all people. Most of all it should not wallow in personal reminiscences, except to provide interesting insights of the beginning of the Society unknown to more recent members. If possible, it ought to be fun without being laboriously funny. The only jokes should be about true events that the whole audience can enjoy. Such events reflect the human side of Science and of ISEK, which I believe is one of those rare pivotal societies where biomedical sciences derive and exchange a great source of energy and substance.

ISEK, of course, was the happy acronym for International Society of EMG Kinesiology, but the rapid broadening of interests changed the EMG in the title to Electrophysiological, an even happier idea. ISEK could just as well represent: 'inspiring science of emerging knowledge,' for that is what it has become. Its practitioners, both in and out of the membership rolls of ISEK, are generally not a narrow-focus group of scientists. Coming from many disciplines, they use ISEK as a meeting place where many bridges meet. This pivotal role remains the underlying principle and purpose for most members of the Society. Looking into the cloudy glass of the future, we can only see a growth in interest and in service that ISEK can expect. One of its special charms is the optimal size of its membership. Neither a great elephant or a tiny flea, it is attractive to many because it permits an international exchange of ideas at an intimate and personal level.

Friendships made at ISEK have lasted for many years and they will endure for many more. The changing topics of special interest will be shaped by the constant shift of interest, but another feature of this Society is the opportunity it gives for both new and old interests to be heard in the same forum, each enhancing the other.

Tomorrow, August 13, 1990, at noon, marks the exact 25th anniversary date of the beginnings of ISEK. At the International Congress of Anatomy in the Rhein-Main-Halle, Wiesbaden, Germany during the summer of 1965, several anatomists with a direct interest in electromyographic kinesiology gathered for luncheon to discuss the organizing of a small society. They were J.V. Basmajian of Canada, S. Carlsöö and B. Jonsson of Sweden, M.A. MacConaill of Ireland, and J. Pauly and

L. Scheving of the U.S.A. Thus, on August 13, 1965, this group agreed that an international society of EMG kinesiology was not only possible but had already begun there and then. Larry Scheving snapped a picture of the birth of ISEK which appeared on the cover of the June 1973 Newsletter.

In 1965 I had the most complete list of potential members, due to my extensive international correspondence leading up to the publication of *Muscles Alive* in 1962. The others were glad to have a volunteer to organize a list of 'charter members from around the world'. With the help of the founders, particularly Carlsöö, Jonsson, and Pauly, by 1967 my list had grown to 231 scientists in 21 countries. This is a good opportunity to give younger and newer members of ISEK an opportunity to know a little bit more about the founding group.

We were, indeed, all anatomists by training and electrophysiologists by choice at that time. Having seen *dynamic anatomy* as the answer to the ills that were besetting the field of gross anatomy, individually we had recognized the pivotal role and utility of electromyography combined with motion studies in restoring the lustre of the ancient science. More important, we had become fascinated by the subject. Let us consider each of the founders in alphabetic order.

I, John Basmajian, M.D., had begun EMG and general electrophysiologic studies in Toronto in 1949 as part of a poliomyelitis clinic at the Hospital for Sick Children. Lacking a teacher, I had assembled EMG equipment based on a clinical device that had emerged from a project in the Canadian Army Medical Corps. With this equipment, I conducted diagnostic examinations of children's motor systems and eventually had a major EMG Clinic going one day a week. A sabattical in London to work with Philippe Bauwens, then the premier clinical electrodiagnostician in the world, confirmed my inclination to use EMGs as both a scientific tool and a diagnostic tool. My spectrum of books eventually confirmed that choice.

Sven Carlsöö, M.D. had acquired a substantial reputation as an EMG scientist and anatomist at the Karolinska Institutet in Stockholm. His early work on neural coordination and biomechanics of mandibular elevation produced one of the classics of electrophysiologic kinesiology, published in 1952. His careful studies combining EMG with other electronic and mechanical instruments over the next two decades was beautifully summarized in his other classic, the book *How Man Moves*, the translation of which appeared in English in 1973. Like myself, he was one of the older men at the founding and his contributions to the literature consequently have tapered off in recent years.

Bengt Jonsson, M.D. was a 'young buck' in the mid-1960s. A member of the Anatomy Department of Gothenberg University School of Medicine, he was just launching a series of major studies with his colleagues which would make him an international figure and lead to his assuming responsibilities in Umea. His work in occupational kinesiology is both innovative and highly practical. Needless to say in an ISEK congress, Bengt also devoted thousands of hours in the past 25 years to this society, of which he was the first secretary when we became more formal in 1967 and had an election.

M.A. MacConaill, M.D. was the pure biomechanist/kinesiologist at that first meeting. He and I were friends largely because I recognized his genius and he loved me for it. Most British anatomists in 1965 were puzzled by this elfin half physician, half engineer, half anatomist – yes he was a 150% man. One of my great conceits is that I helped to get his work accepted through our joint publications. 'Mac' died two years ago in his birthplace of Cork, Ireland. In ISEK he was more a spectator in later years, but I wish all of you had known this scientific brain-trust.

John Pauly, M.D. was, and is, another kind of genius. In the 1950s and 1960s, his EMG studies of respiration with David Jones and Larry Scheving turned that field upside-down. Jones drifted into other fields and Scheving (one of our founders) practically created chronobiology. John Pauly was seduced by that marvellous new field and together they have become world famous. He is now vice-chancellor at the Medical University of Arkansas in Little Rock.

Larry Scheving, who took that archival photograph at Wiesbaden, caught the chronobiology madness quite early and he still remains in a frenzy of his studies in Little Rock though he has passed his 70th birthday.

Now let us return to the development of ISEK.

The Second Period

During the Biomechanics Symposium in Zurich August 22, 1967, some 25 members were available for a business meeting. They decided on a more formal structure and authorized and interim executive consisting of Basmajian (as president), Jonsson (as secretary), and Carlsöö (as treasurer). The executive also included J.E. Pauly, M. Hebbelink and T. Tokazane. They were asked to organize both a first international meeting in Montreal in 1968 and to prepare plans for a formal society. Later Thérèse Simard was recruited to assist as local program organizer. After his move to Montreal, Herb Ladd also plunged into the work of the conference.

The 1st International Meeting of Electromyographic Kinesiology took place in the Queen Elizabeth Hotel in Montreal, August 24–25, 1968, and involved approximately 70 participants. During the two days, 30 scientific papers were presented. Most of them were published as Proceedings of the Meeting in a separate volume of the journal *Electromyography.* You will recognize some of the names of the authors:

Berttioz and Wisner
Biggs and Blanton
Bouisset and Goubel
Brandell
Carlsöö and Schwieler
Cunningham and Basmajian
Hebbelinck
Ikai
Jansen
Jonsson
Jorgens und Hormann
Joseph
Khan and Forrest
Ladd and Smith
Laville and Wisner
Lindberg-Broman, Jonsson & Brolin
O'Connell
Oliveros and Ortiz
Pruzansky
Sack and Muss
Serra
Shipp
Simard and Ladd
Vissner and Bernsten
Weathersby

The subjects were as varied as those on the program of the 25th Anniversary Congress gathered here in Baltimore.

On August 25, 1968, the first official Business Meeting of ISEK was held with about 50 members present. The following subjects were discussed during that Meeting.

1. Election of the councillors was made by voting on ballots distributed to all members and received by the secretary of the Interim Executive before the Business Meeting.

2. The 'Constitution and Rules of ISEK' were adopted at the meeting.

3. It was suggested by William D. McLeod that ISEK should organize a 'Committee for Standards and Definitions'. The proposer was asked to be the chairman of such a committee.

4. Erwin R. Tichauer suggested that ISEK should try to cooperate with other organizations relating to electromyography, biomechanics and kinesiology. The suggestion was unanimously adopted.

The first regularly elected council of ISEK (for four years) were: J.V. Basmajian (president), J. Joseph (vice president), B. Jonsson (secretary), S. Carlsöö (treasurer), F. Buchthal (Copenhagen), V. Janda (Prague), J.E. Pauly (Little Rock, U.S.A.), N.E.J. Rosselle (Publications Secretary) (Louvain, Belgium), T. Tokizane (Tokyo).

The Maturing Years

In 1968 the first ISEK Newsletter appeared. It was edited by B. Jonsson. The above mentioned committee on standards and definitions with W.D. McLeod as chairman was finally formed in 1970, and reported their suggestions in ISEK Newsletter No. 10, 1970. This prompted vigorous discussion in succeeding Newsletters.

In 1972 the second council was also elected. It consisted of I. Petersén (President), O. Lippold (Vice President), H. Ladd (Secretary), T. Simard (Treasurer), J. Basmajian, S. Bouisset, J.E. Desmedt, A.L. O'Connell and P. Pinelli. The 2nd International Meeting of ISEK organized by J. Basmajian and B. Jonsson was held in Barcelona, Spain, July 2-6, 1973, as part of the 6th International Congress of Physical Medicine.

The 3rd International Meeting of ISEK was held in Pavia, from August 30–September 9, 1976. The secretary general of the meeting was P. Pinelli of Italy. A third council of ISEK was elected to serve during the period of 1976–1980. It consisted of H.W. Ladd (President), J.R. Silver (Vice President), G. Andersson (Secretary), H. Broman (Treasurer), I. Petersén, J. Desmedt, B. Jonsson, P. Pinelli, T. Simard, and D.A. Winter.

In the summer of 1979 the members met again for the 4th International Congress of ISEK, this time in Boston (Aug. 5-10, 1979). The secretary general of the meeting was C.J. De Luca. A book containing the proceedings of that meeting was published.

The 5th International Congress of ISEK was held in Lujbljana, Yugoslavia, from June 21-24, 1982, with F. Gracanin as Secretary General.

A new ISEK Council was elected by mail ballot in October, 1984. The Council included: G. Andersson (President), C. De Luca (Vice President), D. Hobart (Secretary), K. Boon (Treasurer), H. Broman, R. Lehr, K. Robinson, Y. Shirai.

The 6th International Congress of ISEK was held from August 26-29, 1985 in Tokyo, Japan. Dr. Tadaatsu Ito, Professor Emeritus of the Nippon Medical School, was the congress chairmån. Fifty-five papers were presented by scientists from 22 nations. Seven major sessions of EMG research were conducted, including: Motor control, Locomotion, Analytical methods, Neurophysiology, EMG methodology, Clinical EMG and Posture.

The 7th International Congress (1988) at the University of Twente, The Netherlands was again a great success. The scientific sessions included 141 presentations, a far cry from the 30 papers at the First Congress in Montreal. I am impressed by the determination of the current Executive and Council to increase the membership well over the figure of 500 which has been our limit. Also the appearance of the *Journal of EMG and Kinesiology* under the editorship of Carlo De Luca is the realization of a marvellous dream by the current Council, who deserve our warmest congratulations and thanks.

Of course there have also been some marvellous Regional Meetings of ISEK since the first one in Umea, Sweden in 1973. These include 1978 meetings in Baltimore; Dubrovnik, Yugoslavia; Monza, Italy; 1981 Tokyo; 1987 here in Baltimore.

Parallel Developments

While ISEK was growing to its present maturity, other societies and their congressess were flourishing too. Through the efforts of Richard Nelson of Pennsylvania State University, Marlene Adrian of Washington State University and Jurg Wartenweiler of Zurich, the *International Soceity of Biomechanics* was founded in 1973, with our considerable sympathy and support. Somewhat timid about the word 'Kinesiology', they elected to go for 'Biomechanics', whereupon they offended many orthopaedists and others who felt they held a patent on that word and work. Admittedly, ISB has a strong physical education spirit but it has been a good sister society for ISEK with many members crossing over freely. I was proud to be a founding member and councillor of ISB for the first three years. Their last congress that I attended several years ago was very successful, but again, one would note the heavy bias toward schools of P. & H.E. in the presentations. I believe ISEK is much more eclectic than ISB.

Where Have All the Anatomists Gone?

In view of its beginnings, one would imagine ISEK would be dominated by anatomists. This is certainly not the case. Moreover, at the latest American Association of Anatomists Annual Meeting in Philadelphia there were only two papers that might interest ISEK members. Twenty-five years ago there might have been 25 of them. Did EMG kinesiology act as a tonic for gross anatomy? Yes, but there were other parallel developments including the formation of the *Society of Neuroscience* which now has 10,000 members. Bizarre as it may seem to you, I was a founding member of that too. My only apology is for its huge bloated size. I haven't gone to any meetings for years. All the more reason for the success of smaller congresses like ISEK's. Of course we should enlarge the paying memberships and support our journal but God forbid that our meetings bulge like the Neuroscience ones.

Other groups we both relate to and compete with are the Biomedical Engineers, the Orthopaedists, the Physical Medicine community, the Physical Therapists, and the EMG Biofeedback research and therapy associations. None of these are threatening and some are friendly siblings. From their memberships we could attract small trickles of members, but not a torrent. Again, we already have small cross-overs, especially with biomedical engineers, witness, among others, our president Carlo De Luca and our council member Moshe Solomonow.

Spin-Offs of ISEK and EMG

Twenty-five years ago, none of us could have imagined that basic EMG would spin off into other broad fields of investigation and clinical applications where it would underpin their methodologies significantly.

General Zoology, an early spin-off was in investigation of other species, which, in retrospect now, seems to have been inevitable. Although zoologists and primatologists have not joined ISEK in large numbers, their work on motor control mechanisms range from the gills of fishes and middle-ear muscles of bats in flight, to the limb-muscle patterns during various modes of terrestrial and aerial locomotion of many species. How well I remember my own introduction to the EMG of other animals at Queen's University in the late 1950s. With Bill Boyd (my PhD graduate student and now a distinguished scientist at Guelph University), we did many exciting studies with multiple and long term electrodes implanted on the diaphragm of rabbits. Other studies of urinary muscle control, of goat embryos and of porcupines followed.

My most interesting one-day study of another species occurred when the famous zoologist Carl Gans, then at Buffalo and later, chairman of the University of Michigan Biology Department, came to see us. He too wished to use EMG in other species and wanted to learn the technique of our inserted fine-wire electrodes. After seeing a demonstration on a human subject (probably himself), he asked if we could do an animal experiment on one of his animals in my office. I was puzzled because there was no evidence of his bringing one in. Opening his brief-case, he brought forth a beautiful 20-cm snake and we spent a marvellous afternoon in studying its movements. After that, Dr. Gans was 'hooked' on bipolar fine-wire electrodes and on EMG, going on to write the excellent book 'Electromyography for Experimentalists' with Gerald Loeb, published by the University of Chicago Press in 1986.

Primatology. Physical anthropologists, many of whom are also primatologists deeply concerned with evolutionary theories, outnumber specialists who have interest only in non-human primates. My involvement at the Yerkes Regional Primate Center of Emory University with the great apes – gorillas, chimpanzees and orangutans – was created by Russell Tattle of the Anthropology Department at the University of Chicago, as I was moving to Emory University in Atlanta. We spent five happy years in intense study of these animals and published some sixteen scientific contributions over a period of a dozen years. Since almost all of these appeared in anthropology journals and books, most of you in this audience would not have even heard of them. This is now typical of the pervasive spread of our discipline of electrophysiological kinesiology. Many people in other scientific fields use the techniques freely without a direct link to this Mother Society.

Even the Arts are not immune. My three PhD graduates in the field of musical performance are a harmonious example of that. In the past few years, clinical electromyographers, too, have invaded the area of problems in performance by musicians, and the use of EMG feedback for treatment.

EMG Biofeedback. The largest spin-off of this society's concerns occurred when psychologists seized my studies of single motor-unit control mechanisms and employed it for relaxation therapy. I happily went along with parallel developments in rehabilitation medicine. The 3000-member Association of Applied Psychophysiology and Biofeedback, which grew from the Biofeedback Society of America and scores of other smaller biofeedback societies around the world, attest to the widespread use of EMG. Some 3000 journal articles and 100 books have been written on biofeedback, 90% of which are EMG feedback. I doubt that any or many members of that neighbouring field are any more knowledgeable about ISEK than you are about them. Nevertheless, they owe a solid debt to this Mother Society and its members, and on every occasion possible, I try to make that point to them for I was a founding member of the original Biofeedback Society too.

The Future

Our growth to maturity has replicated the usual life of a person to the present. Conceived by accident and nurtured by a tiny family for several years, ISEK managed to survive until its christening at age 3. With a diploma of legitimaticy, i.e., its Constitution, and elected guardians, i.e., the Executive, it's led a fairly good growth pattern. Today at 25, it has graduated from its postgraduate training and it holds its PhD – 'Pretty Hot Discipline'. Will it also acquire higher honours? Time will tell. Will it sicken and decline? Almost certainly not if one is to judge by its sober growth. I use the word sober for the Society's growth, not for the behaviour of some of its members at parties that has been anything but sober.

In short, I am very optimistic about ISEK, expecting it to outlive even the youngest current member. As long as it serves its members well it will thrive. If, however, at some distant time interest in zoologic control of propulsion and posture, especially in human beings, dies then ISEK will die too. Who in this room can imagine the former? I cannot. So I see a flourishing adult life for ISEK. *Long Live the International Society of Electrophysiologic Kinesiology.*

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ELECTROMYOGRAPHY

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

METAMORPHOSIS OF REHABILITATION FROM ART TO SCIENCE: THE ROLE OF QUANTIFIED ELECTROMYOGRAPHY

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For over 60 years, scientists and clinicians have steadily advanced their understanding of Sherrington's "final common pathway" through the study of muscle properties and behaviors as revealed by electromyography. The sophistication with which we are able to make qualitative and quantitative judgments about how muscles work has advanced at almost an exponential rate. Indeed, the integration of electromyography with techniques designed to comprehend, among other things, motoneuronal behavior, evoked responses within the central nervous system, histochemical profiling, axonal uptake and retrograde transport of selective dyes, and muscle metabolism through glycogen depletion techniques have heralded a productive age of investigation into the makeup and mechanics of muscle on the one hand and their control over movement on the other.

While our basic understanding of muscle has progressed with the development of tools to better record, store and analyze electromyographic (EMG) activity, the advances of clinical EMG have not necessarily followed a parallel course, especially within the field of rehabilitation. Of course, clinicians can make excellent use of percutaneous recordings and their storage to differentially determine pathologies of neurogenic or myogenic origin. Electromyography, along with kinematic analyses, have been the mainstay of kinesiology. Collectively, these scientific endeavors have led to a better understanding of the circumstances under which muscles function.[1-3] But to what extent has this knowledge base been integrated into clinical practice? How, for instance, does kinesiological assessment play a role in clinically determining the extent to which patients with weakness or movement control limitations improve? Regrettably, the answers to these questions have not been pursued rigorously, until recently.

Reality dictates that most therapeutic interventions of long-term consequence for the restoration of movement control and function are performed by health-related professionals. For decades, the primary, and at times sole proprietary tool, of these practitioners was their hands. Most physical rehabilitation providers take great pride in the skills they have acquired in perfecting the <u>art</u> of their practice and will proudly proclaim the invaluable gift of their hands and observational talents; for <u>touch</u> and <u>vision</u> have led many a clinician to make remarkable improvements in a patient's status. The difficulty with the perfection of these arts resides in the growing inappropriateness, within today's competitive market, for reimbursement of medical and rehabilitative services. For unfortunately, while there will always be a need for the art of touch and the interpretation of palpatory observation, vision and touch are simply not amenable to meaningful interpretation or quantification; and it is quantification of function and all physiological factors contributing to an understanding of functional potential that are forming the bases for decisions regarding continuance of treatment and grounds for reimbursement of services.

The need for vehicles to transform the art of rehabilitation into a science is clearer now than ever. In the realm of areas such as cardiac or pulmonary rehabilitation, the waters are far less murky; for clinical and laboratory findings, such as blood and gaseous measures, help to validate the value of rehabilitation services. Those of us concerned with movement dysfunction have a quantitative tool in EMG which is only now beginning to show valuable contributions to the documentation of physiotherapeutic interventions.

Ever since the pioneering studies of Basmajian and his colleagues[4] in the area of single motor unit control, we have known that man is capable of modifying muscle activity if the appropriate audio and visual interfaces are provided. The success of EMG biofeedback applications (as reviewed by Basmajian[5] and Hatch, Fisher and Rugh[6]) has hinged on the timeliness of feedback signals and the specificity of recordings from electrode placements.[7] This modality is widely used in the United States and elsewhere, but, one might contend, in a less than contemporary manner.

The notion of a relatively isolated machine-human interface still permeates the thoughts of many clinicians. Basically this belief resides in the connection of patient to machine, and manipulation of the latter by the former with very little guidance from the clinician. Not until 1982 did we come to realize that perhaps the feedback was as valuable to the clinician as it might be for the patient.[8] For indeed, if the clinician would take the time to integrate the unique palpatory skills with the information provided to him from the feedback derived from the muscles of patients, the finely graded changes in hand positioning, resistance to movement, or patient positioning could be facilitated. The feedback device can serve as a first approximation of what the underlying muscles are doing in response to a verbal command or physical manipulation. Sadly, Too many clinical specialists are so ingrained with an addiction to watch the patient at all costs that the information transduced in front of their eyes cannot be heeded. The ability to acquire the skill of handling the client while observing the visual representations of EMG emanating from one or more channels of processed activity provides an integral link in the metamorphosis

of rehabilitation from art to science. For the scenario just described is probably an example of applied electromyography or, in a more refined sense, when movement changes are also taken into account, applied kinesiology. Truly this blend of palpation and observation and utilization of processed biological signals can assist the clinician to better comprehend patient responsiveness to commands or to the application of any of a number of neuromuscular re-education techniques. The issue no longer becomes one of whether a certain approach to treatment is effective, but rather, one of "on line" modification of the approach to achieve the desired response.

What is now needed as an addendum to this interface is a way of quantifying EMG responses for defined time intervals corresponding to the movement, exercise, or task commanded by the clinician. With contemporary microprocessing capabilities, this concern is really trivial. We refer to the on-line quantification of EMG activity as concurrent assessment of muscle activity or CAMA.[9] Simply by successively activating a remote toggle switch to any EMG processing device, one can delineate and extract average and peak values of EMG defined over that interval. In this manner the clinician and patient have full view of the feedback signal during the interval and documentation at its conclusion. As an example, one might take the activity of "bridging" as a strategy designed to recruit the hamstring muscles as hip extensors. In this task, the patient is supine with knees bent and is instructed to push the foot into the mat and raise the buttocks. Relevant information that could be assessed through this on-line EMG feedback analysis includes the appropriate hip and knee joint angles, as well as the best contact surface against which to push to achieve the best hamstring response. To date, this level of sophistication in explaining many a therapeutic procedure has been absent. By combining these data with other measures related to, in this example, relevant features of gait, such as cadence, velocity, numbers and types of assistive ambulatory devices, etc., one begins to amass a quantitative scheme that places treatment, its duration and its outcomes in a more defensible position for reimbursement and for comprehending the bases of the treatment undertaken. From a scientific perspective, this approach provides a myriad of opportunities for clinically researchable questions and, more importantly, opens the practice of rehabilitation to more careful and deserving scrutiny.

The limitations of this form of quantification, however, are inherently the limitations of surface electromyography. Do the processed data accurately reflect the activity of targeted muscles underlying the electrode placement? Is the response so recorded truly related to functional consequences? Is the variability in responses from spastic or rigid muscles so great within and between individuals as to make the value of this approach meaningless when applied to such muscles? Is EMG the best or most appropriate biological signal to be transduced and quantified? These are important questions that must be answered.

But in the interim, within the context of physical rehabilitation, one must realize that EMG and its interpretation has been an integral part of diagnosis, not of clinical treatment. In this regard, we must encourage the use of EMG as an evaluation and treatment tool designed to foster assessment and promote the inquiry and solution to researchable questions. By acquiring this posture we can provide one small contribution to the scientific development of rehabilitation as this century comes to a close.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

VALIDITY OF THE INITIAL CALIBRATION CONTRACTION IN STUDIES OF LOCALIZED MUSCLE FATIGUE: AN EMG STUDY.

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INTRODUCTION

In 1962 Kogi and Hakamada showed that the frequency spectrum of the EMG shifted towards lower frequencies in sustained muscular contractions. This observation has been used in many situations to evaluate muscle fatigue. The decrease of frequency is often calculated as a quotient between the actual frequency and the initial frequency, obtained by a calibration contraction. Our research group has examined different kinds of variation of MPF in the trapezius muscle (Öberg et al 1990 a,b).

A critical question concerning the frequency shift method must be: Is the initial calibration value valid for the whole range of movement and for different load situations? The aim of this study is to describe the systematic variation of MPF in relation to a calibration value obtained by a test contraction in a predetermined joint position. This variation can be used to evaluate the validity of the test contraction.

MATERIAL AND METHODS

19 healthy people were examined. The study was limited to the functional range of movement in the shoulder joint (0-135 degrees, flexion and abduction in the scapular plane) and with hand loads corresponding to the weight of common hand tools (0-2 kg). Surface and intramuscular EMG were recorded from the trapezius muscle The signal was amplified, tape recorded, A/D-converted, analyzed in a computer, FFT-analysis was performed and mean power frequency was calculated. The obtained values were examined statistically in a multiple regression model with respect to angle, load and computed torque at the shoulder joint. The experimental setup is described in detail elsewhere (Öberg et al 1990a).

RESULTS

The variation of MPF is related to a proposed calibration contraction: straight arm, 90 degree abduction in the scapular plane, 0 kg hand load.

Surface EMG: The results of the multiple regression analysis are shown in table 1. Statistically significant variation was seen only in relation to shoulder joint angle. The magnitude of the variation can be evaluated in the diagram (fig 1): max 8% of the calibration value in the range 30-135 degrees of abduction.

TABLE 1.
REGRESSION ANALYSIS. SURFACE EMG. 0 DEGREES EXCLUDED.

VARIABLE	REGRESSION COEFF x 100	STANDARD ERROR x 100	STUDENT'S t-TEST	P-VALUE
ANGLE	0.13	0.013	10.5	<0.001
LOAD	0.26	0.814	0.3	N.S.
TORQUE	0.20	0.112	0.4	N.S.

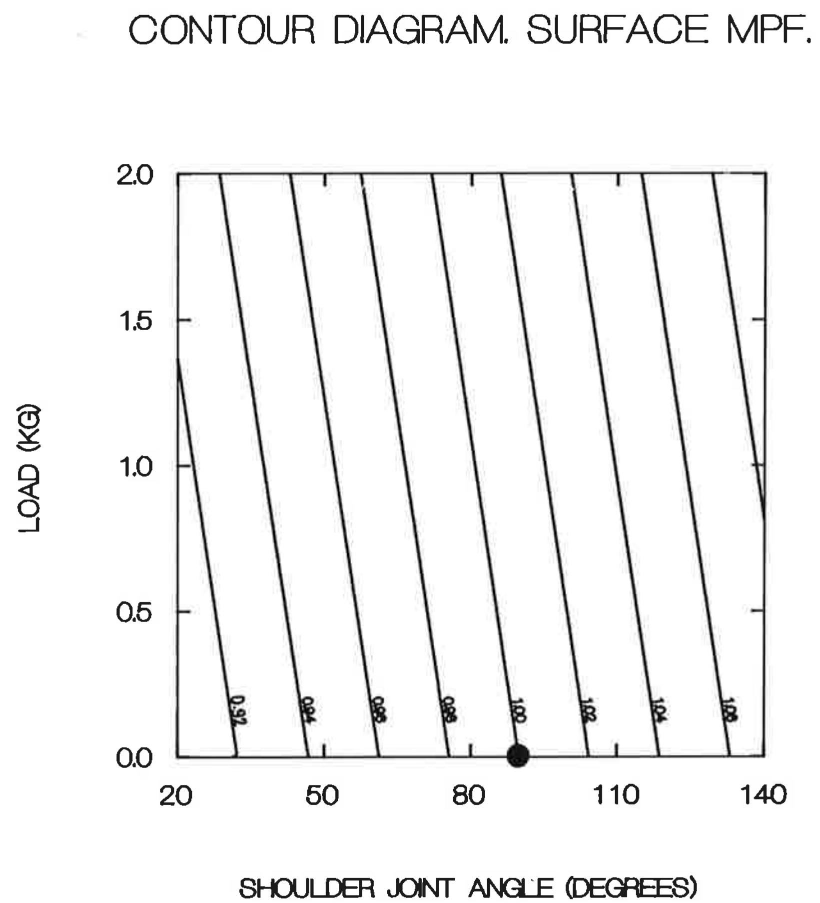

Fig 1. Linear contour diagram showing the variation of MPF in relation to shoulder joint angle and external hand load. The distance between the parallell contour lines is 0.02. The black dot indicates the calibration position. Surface recordings. Zero degrees excluded.

Intramuscular EMG: The results of the multiple regression analysis are shown in table 2. Statistically significant variation was seen only in relation to external hand load. The magnitude of the variation can be evaluated in the diagram (fig 2): max 7% of the calibration value in the range 30-135 degrees of abduction.

TABLE 2.
REGRESSION ANALYSIS. INTRAM. EMG. 0 DEGREES EXCLUDED.

VARIABLE	REGRESSION COEFF x 100	STANDARD ERROR x 100	STUDENT'S t-TEST	P-VALUE
ANGLE	0.03	0.017	1.5	N.S.
LOAD	-3.33	1.086	-3.1	0.002
TORQUE	0.08	0.149	0.5	N.S.

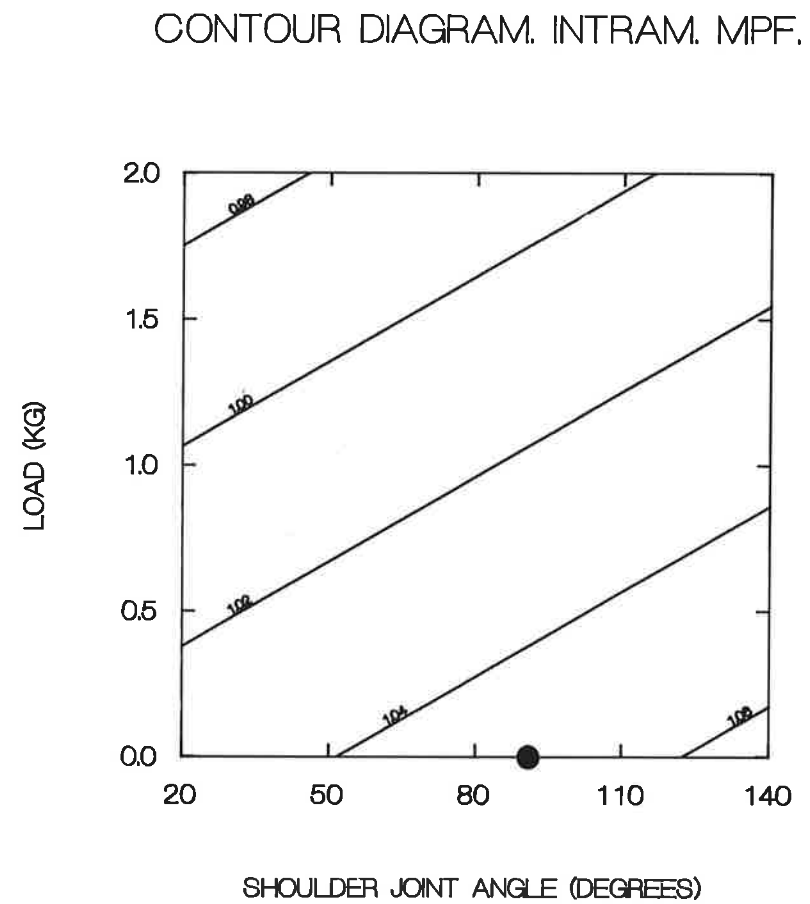

Fig 2. Linear contour diagram showing the variation of MPF in relation to shoulder joint angle and external hand load. The distance between the parallell contour lines is 0.02. The black dot indicates the calibration position. Intramuscular recordings. Zero degrees excluded.

DISCUSSION

The actual study was performed on the trapezius muscle. This muscle is a common site of work related chronic myalgia. It has been proposed that local muscle fatigue can be a causative factor. That is the reason for our choice of muscle. However, we believe that the results of this study could be applied to other muscles as well.

In this study we have shown, that there is a systematic linear variation of MPF that is not related to muscle fatigue, but to other factors such as shoulder joint angle and external hand load. We found a maximal deviation from the proposed calibration position of 8% for surface recordings and 7% for intramuscular recordings. These figures are valid only within the functional range of movement in the shoulder joint (0-135 degrees) and for hand load 0-2 kg or combinations within this angle/load range. At very low signal levels the influence of noise give distorsions in the data analysis. For this reason zero degree joint angle (the arm hanging down) was excluded from the analysis. In a recent study we have shown, that the difference between surface and intramuscular recordings can, at least partly, be explained by a geometric displacement between the surface electrodes and the underlying muscle tissue (Öberg et al 1990b).

We suggest, that the MPF calculated from a proposed calibration position (90 degree abduction, 0 kg hand load) is regarded as valid within the functional range of movement (0-135 degrees) under certain conditions: The change of MPF must exceed 8% for surface recordings and 7% for intramuscular recording to be indicative of local muscle fatigue. To this must be added a certain degree of random variation.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

ON-LINE ESTIMATION OF SURFACE MYOELECTRIC SIGNAL SPECTRAL PARAMETERS: METHODOLOGICAL CONSIDERATIONS

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INTRODUCTION

It is well known and documented that the power spectral density function of the myoelectric signal undergoes frequency compression during either voluntary or electrically elicited sustained contractions [3,4]. The variations of myoelectric signal parameters due to spectral compression are generally referred to as myoelectric manifestations of localized muscle fatigue. Previous studies [1] showed that during electrically elicited contractions median frequency (MDF), mean frequency (MNF) and average rectified value (ARV) are the most sensitive parameters to muscle fatigue. In the cited work [1] it was also found that during voluntary contractions amplitude parameters are not reliable indicators of fatigue.

Spectral parameters may be computed either off-line or on-line. Off-line techniques provide greater accuracy, but for clinical purposes, when a compromise between accuracy and computational time must be sought, on-line techniques may be preferable. Analog on-line computation of MNF and MDF has already been performed by previous researchers [5,7], and some devices suitable for clinical use have been described and validated [5]. Unfortunately, the analog approach requires dedicated circuitry and careful calibration procedures. The availability of DSP-based cards to be used with AT or 386 personal computers and their relatively low cost justify a different approach to spectral parameter estimation. Standard industry cards may be used for performing on-line spectral estimation. The power spectral density function and its related parameters may be computed by means of FFT-based techniques. DSP chip producers offer a variety of different products and dedicated software libraries. Previous authors [2] applied standard industry cards and software libraries to myoelectric signal processing.

The aim of this work is to present a comparison between off-line and on-line techniques applied to myoelectric signals obtained during voluntary and electrically elicited contractions, stressing the attention on some methodological aspects related to FFT algorithm selection. The on-line system was realized by means of a 286 AT-like computer equipped with a TMS 320C25-based board produced by Loughborough Sound Images L.t.d. The effect of finite word length and fixed-point arithmetic on FFT algorithms is considered. Starting from the results reported in this work, an on-line system suitable for clinical use has been developed and is currently being evaluated.

METHODS

Two voluntary and four electrically elicited contractions lasting 25 s were performed with the right tibialis anterior of a healthy human subject. Contraction modalities were chosen based on previous results [1,8].

Voluntary contractions were performed at 20 and 80% MVC. The rationale supporting this choice is twofold: I) the chosen contraction levels cause very different myoelectric manifestations of fatigue [1], and II) the root mean square value of the myoelectric signal is substantially different at the chosen force levels, and its extremes identify the relative range of variability that could be expected in clinical applications. Consequently, both the response of the system to different rates of spectral compression and its sensitivity to numerical overflow could be evaluated.

Electrically elicited contractions were performed with stimulation frequencies of 20 Hz and 40 Hz, and at two different stimulation amplitudes. High level stimulation (HLS) produced a maximal M-wave, low level stimulation (LLS) produced an M-wave about 30% of the maximal one.

The leg and the foot of the subject were placed in a brace to obtain isometric conditions. The knee was extended and the ankle joint fixed at approximately 90°. Electrical stimulation was applied to the main motor point of the investigated muscle with the monopolar technique. The detection probe was placed between the most distal motor point and the tendon [9]. Surface myoelectric signal was detected by means of a differential active electrode (10 *mm* interelectrode distance), amplified, and band-pass fltered. Stimulation and detection circuits have been described in greater detail elsewhere [6].

The myoelectric signal was simultaneously processed by the on-line system and recorded on the hard disk of the off-line one. The time courses of spectral parameters computed on-line and off-line were tabulated, plotted, and compared.

<u>ON-LINE DATA PROCESSING:</u> Signals were sampled at 1024 *Hz* by a 16 bit A/D converter mounted on the DSP board. Spectral parameters were computed starting from raw periodograms obtained by a decimation in time 128-point FFT algorithm. The algorithm chosen was a basic one, in which no measure was taken to control possible numerical overflow [10]. Fixed-point arithmetic with 16-bit wordlength was used, to take full advantage of the TMS 320C25 architecture.

The sequence of samples of the input signal was segmented in strings of 128 samples, from which the power spectrum was estimated by means of the raw periodogram. The values of spectral variables were then computed for each consecutive epoch.

<u>OFF-LINE DATA PROCESSING:</u> Signals were sampled at 1024 *Hz* and converted by a 12 bit A/D card. Floating point arithmetic with 32-bit wordlength was used for the computation of spectral parameters. Double-precision (64-bit wordlength) was used in the trigonometric recursion of the FFT algorithm.

Myoelectric signal power spectrum MNF and MDF were computed over consecutive epochs lasting 1 *s* for both voluntary and electrically elicited contractions. This epoch length is generally short enough for detecting promptly the onset of muscle fatigue [8], yielding in the meantime acceptable errors on the spectral parameter estimates [1].

Power spectra of stimulated contractions were computed from the electrically elicited responses averaged over a 1 *s* signal epoch. Zero padding in time domain was used to interpolate the power spectrum, thus obtaining spectral lines 2 *Hz* apart. To allow for direct comparison of the results obtained on-line and off-line, power spectra were also computed without averaging.

RESULTS AND DISCUSSION

In Fig. 1. the time courses of MDF during voluntary contractions are reported. At 20% MVC there is a good agreement between on-line and off-line estimates. On-line values show a slightly higher bias and coefficient of variation than off-line counterparts. This effect is due to the epoch length (1/8 s on-line vs. 1 s off-line), and was studied in a previous work [1]. At 80% MVC the real time system overestimates MDF and MNF. This is due to numerical overflow in the FFT algorithm. It is well known that FFT algorithms are much more efficient from a computational point of view than the basic DFT, but their higher sensitivity to numerical overflow, especially when fixed-point arithmetic is used, is too often underestimated. Results reported in Fig. 1. demonstrate that FFT algorithms to be used with on-line systems featuring fixed-point arithmetic must be able to control the occurrence of numerical overflow conditions. A detailed discussion of the problem and theoretical and empirical evaluations of different strategies was reported by Welch [10].

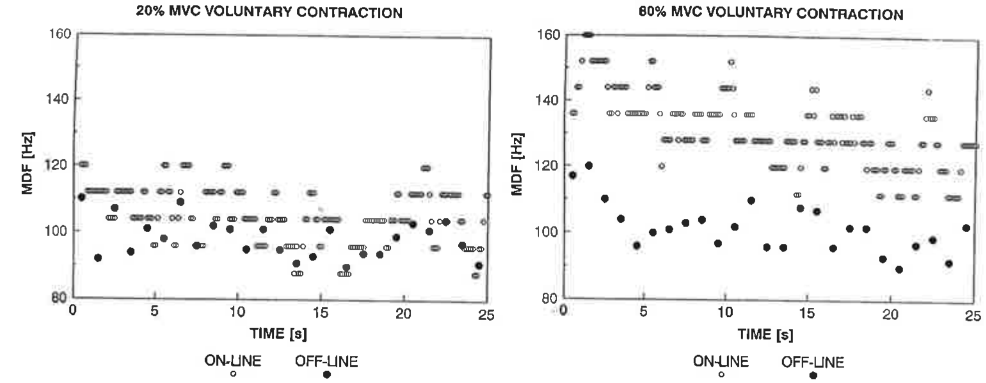

Fig. 1. Time course of median frequency during voluntary (20% MVC and 80% MVC) contractions. On-line and off-line estimation techniques are compared.

Fig. 2. represents MDF and MNF estimates obtained during an electrically elicited HLS contraction. During electrical stimulation the signal is periodic, and consequently its power spectrum is discrete and the distance between spectral lines is equal to the fundamental frequency of the signal (viz., the stimulation frequency). This is evident from the behavior of MDF when estimated off-line and without averaging ((•), spectral resolution equal to 1 *Hz*). In this case MDF is not able to track promptly spectral compression. In the same situation MNF tracks it much more efficiently. To overcome MDF limitations, M-wave averaging and spectral interpolation must be used, as evident from the time course of MDF computed with averaging and interpolation (■).

From the observation of the time course of MNF and MDF estimated by the on-line system (○) it is evident that, in this case, the behavior of MNF and MDF is similar. In fact, because of fixed-point arithmetic, both MDF and MNF are multiples of the distance between consecutive spectral lines. Since the estimated power spectrum is given by the convolution of the real one with the power spectrum of the chosen window, spectral lines 8 *Hz* apart were generated (the main-lobe width of the rectangular window was 8 *Hz*).

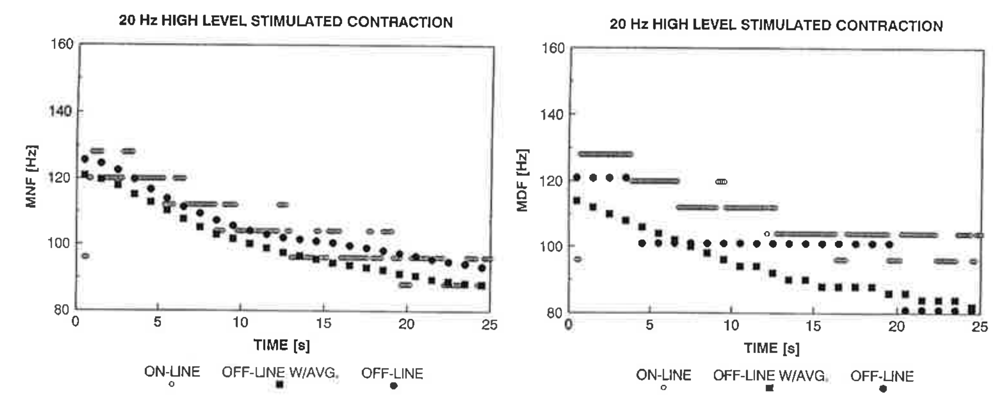

Fig. 2. Time course of median frequency and mean frequency estimated during electrically elicited contractions.

CONCLUSIONS

In this work there are two major findings: 1) to obtain on-line systems able to compute spectral variables of surface myoelectric signal suitable for clinical use, FFT algorithms able to control numerical overflow must be implemented, and 2) when fixed-point arithmetic is used MDF and MNF show a very similar behavior during voluntary and electrically elicited contractions.

ACKNOWLEDGMENT

We express our gratitude to Z. Erim, L. Lo Conte and P. Lovisolo for their help in this study. This work was performed within the framework of a scientific cooperation between the Dept. of Electronics of Politecnico di Torino and the Dept. INFOCOM of Universita' di Roma, La Sapienza. Support was provided by the Italian Ministry for University and Scientific Research and by the National Research Council of Italy.

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SIGNIFICANCE OF TEMPERATURE CHANGES IN VOCATIONAL EMG

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INTRODUCTION

In vocational studies EMG is often used to quantify muscle activity (5, 6, 12, 14). Changes in muscle temperature (T_m) may, however, change the EMG signal without any change in muscle activity. The mean power frequency (MPF) is generally agreed to decrease with a reduction in T_m (e.g. 2). In contrast, different results have been reported concerning the direction and quantity of amplitude response to temperature changes (9-11).

The present data quantify the effect of ambient temperature on EMG amplitude during dynamic contractions within a temperature range which may occur in vocational EMG studies.

MATERIAL AND METHODS

Six healthy male students (age: 23-35 years, stature: 168-185 cm, weight: 62-82 kg) volunteered. EMG was recorded from the lateral part of the left soleus muscle by a pair of surface electrodes. Skin temperature (T_{sk}) close to the electrodes was recorded every 15 min.

After 2 hours of relaxed sitting at an ambient temperature (T_{amb}) of 30°C, the subjects performed bilateral heel lifting (8 cm, 24 min^{-1}) for 10 minutes. This procedure was repeated at 14°C.

During the leg activity the EMG signals were recorded on tape. The EMG analyses included calculation of the amplitude probability distribution function (APDF) and the MPF. Details concerning instrumentation and analyses of the EMG signals were as previously described (4, 5, 13, 15).

RESULTS AND DISCUSSION

The mean T_{sk} during the EMG recordings was 32.9°C in the warm condition and 21.7°C in the cold condition. Almost the same reduction in T_{sk} may occur during an 8 h working day in a seated position at an ambient temperature of 21°C (9), and lower leg T_m may decrease by about 4°C (15).

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The cooling of the lower leg led to an amplitude increase as illustrated in Fig. 1. For the probability level (P) 0.9 the amplitude increased from 73μV to 135μV (p=0.02, 2-tailed).

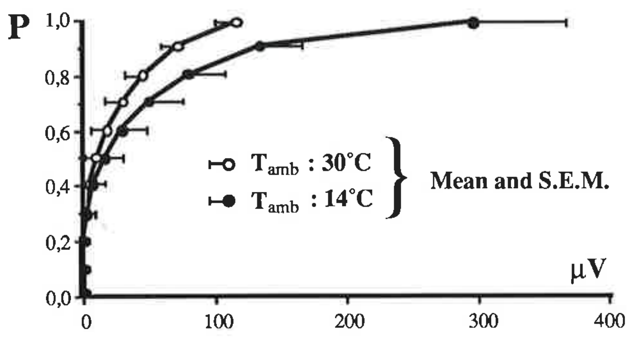

Fig. 1. Mean APDF´s of the soleus muscle during the exercise A3. T_{amb}: ambient temperature. Further details in text.

The resting EMG level showed a non-significant difference between the warm (4μV) and the cold condition (7μV). Thus, the amplitude increase was not due to shivering.

However, it may partly be due to the simultaneous reduction in the MPF (from 142Hz to 83Hz in average, p=0.004, 2-tailed) in combination with the low-pass filtering properties of the tissues (7); this allows more signal energy to pass through from the active muscle fibers to the electrodes (1).

In addition, increased viscosity of muscle tissue with reduced temperature may lead to a higher loss of force during dynamic than isometric contractions. This may explain the difference between our results and those of Petrofsky (9) and Petrofsky & Lind (10) who studied isometric contractions.

CONCLUSIONS

The EMG amplitude of the soleus muscle during modest dynamic exercise (concentric contractions) is approximately doubled when the T_{sk} above the muscle is reduced from about 33°C to 22°C during a 2 hour period. Thus, T_{sk} should be carefully controlled in vocational EMG studies.

ACKNOWLEDGEMENTS

This investigation was supported by grants from the Swedish National Board of Occupational Safety and Health, Nordiska Forskarkurser (No. 86/87) and The Norwegian Academy of Science and Letters (No. 31873). We would like to thank Henrik Klitgaard and Lise Riis for valuable assistance in the experimental work.

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COMPARISON BETWEEN MUSCLE FIBER CONDUCTION VELOCITY ESTIMATION TECHNIQUES: SPECTRAL MATCHING VERSUS CROSS-CORRELATION

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INTRODUCTION

Muscle fiber conduction velocity (CV) is a basic physiological parameter related to the excitation properties of the muscle fiber membrane. The aim of this work is to compare two broadly accepted and used CV estimation techniques: *Cross-Correlation* (CC) [5,8,6] and *Spectral Matching* (SPM) [3,2,4]. In both cases CV is estimated by computing the time delay τ between myoelectric signals detected by means of array electrodes at two different sites (d mm apart) along the muscle fibers. Starting from the delay τ, CV may be computed by means of the well known relationship $CV\ [m/s] = d\ [m]/\tau\ [s]$.

The two aforementioned methods are theoretically equivalent, as they both compute the time delay between two signals by shifting one signal with respect to the other until the mean square error (MSE) between them is minimized. When the CC approach is used, the delay that minimizes the MSE is estimated by localizing the maximum of the cross-correlation function. The SPM technique relies on the fact that aligning two different signals in the time domain according to the LSE criterion is equivalent to minimizing the MSE between their Fourier transforms. The estimation of the delay between two signals in the frequency domain has been discussed in detail by Dorfman and McGill [3].

In order to achieve an acceptable resolution ($\pm 0.5\%$ of typical CV values) in CV estimates, signals must be oversampled or the cross-correlation function must be interpolated. Roy et al. [6] described a possible interpolation algorithm applied to the cross-correlation function. Other interpolation techniques have been presented in literature. Previous authors [3] discussed the computational advantages of the SPM technique. A thorough comparison between the two techniques applied to the non-invasive estimation of muscle fiber CV has not yet been presented. In this work we discuss the sensitivity of conduction velocity estimation performed by means of the CC and SPM techniques to added white noise, different CV values, and quantization errors due to the A/D conversion.

MATERIALS AND METHODS

The two signals to be aligned were synthesized by means of a new computer package [1] able to simulate either single or double-differential signals detected by means of array electrodes. The simulated myoelectric signals consisted of 25 active motor units (MU). Their firing rate ranged from 18 to 35 pps and the IPI standard deviation from 3 to 16%. The CV of the active MUs was kept constant throughout each contraction. Signals were generated with a sampling rate of 1024 Hz. Each simulated contraction lasted 10 s and, for computational purposes, it was subdivided into 20 epochs lasting 500 ms each. CV values obtained with CC and SPM techniques were computed for each epoch, thus obtaining a time series of 20 values for each simulated signal and estimation technique.

The bias and the variation coefficient of the estimates were used to characterize the estimation errors of the two different techniques, whose performances were then compared by means of the paired Wilcoxon test [7].

EFFECT OF ADDITIVE GAUSSIAN WHITE NOISE

The study was performed for superficial and deep muscle fibers. The depth of superficial muscle fibers ranged from 1.5 to 5 mm, while that of deep muscle fibers ranged from 7 to

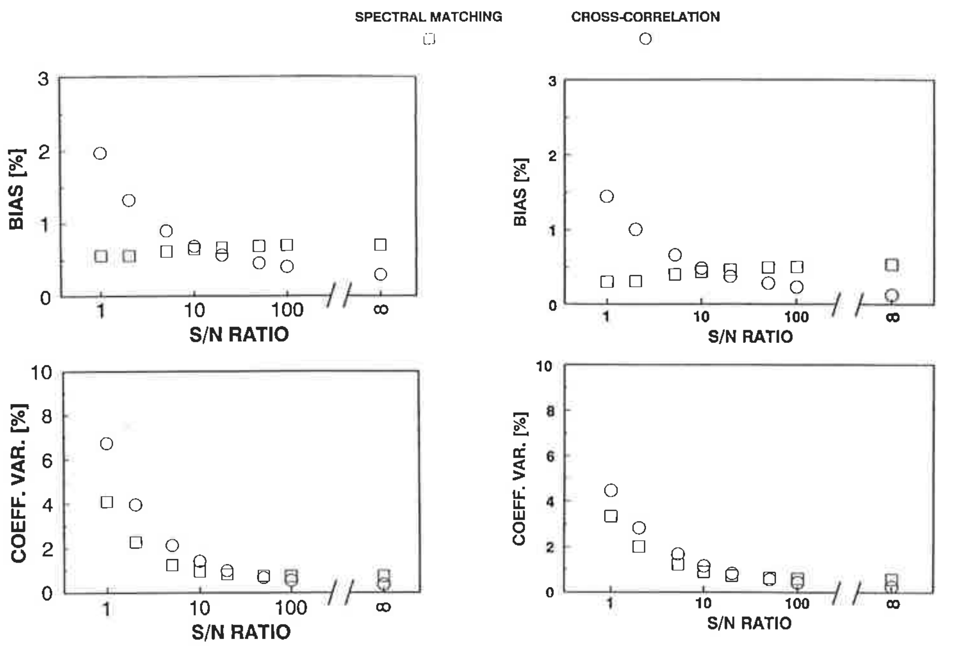

Fig. 1. Sensitivity to additive Gaussian white noise for deep (left) and superficial fibers (right). Values are reported as percent of the real value.

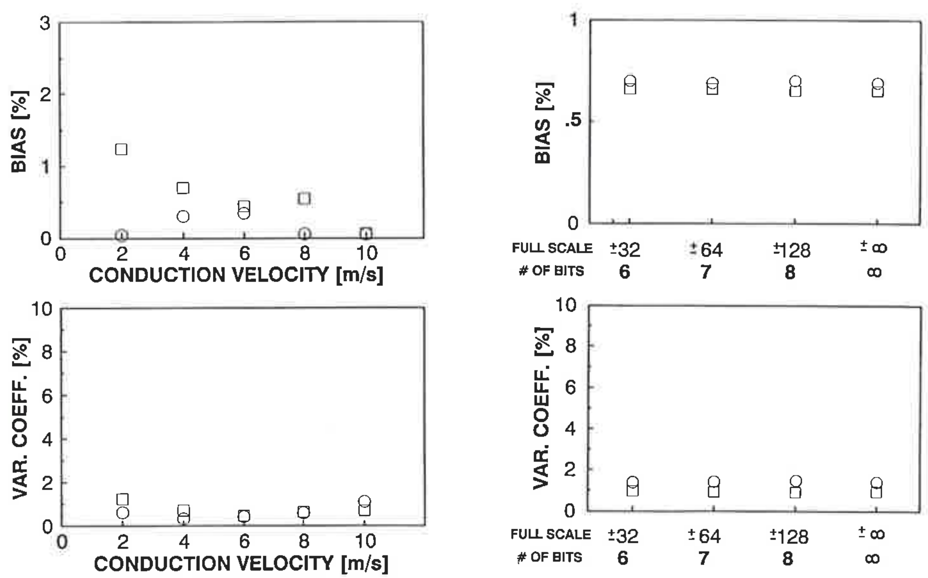

Fig. 2. (A) Effect of different values of conduction velocity. (B) Effect of quantization error.

11 mm. In both cases active MUs had a constant conduction velocity equal to 4 m/s. First the signals were generated without any noise contribution, then the average power of the signal in each epoch was computed and Gaussian white noise was added to the signal. S/N ratios ranging from 1 to infinity (1,2,5,10,20,50,100,∞) were obtained. Bias and coefficient of variation of CV estimates were computed for each condition, tabulated, and plotted.

Effect of Different Conduction Velocity Values

CV affects the shape of the power spectrum of the myoelectric signal and consequently also the performances of the estimation techniques. We evaluated the performances of CC and SPM for different values of CV, ranging from 2 to 10 m/s. The simulated myoelectric signal consisted of 25 MUs whose depths ranged from 7 to 11 mm. CV was kept constant during each simulated contraction and five different CV values were chosen (2,4,6,8,10 m/s). No noise was added to the signals. Again, bias and coefficient of variation of the estimates were computed for each CV value, tabulated and plotted.

Effect of Quantization Error

Since the application of CV estimation in the clinical environment is raising increasing interest (see [9,10] among others), and since simple and inexpensive devices must be built in order to allow physicians to take advantage of the information content of the CV time course, it is important to investigate the effect of the quantization error on the performances of CV estimation techniques.

The same simulated signal used for investigating the effect of additive noise was used for this study. Starting from it, different resolutions of the A/D converter (infinite, 8, 7, and 6 bit) were simulated to generate the signals then used to estimate the CV. Bias and coefficient of variation of the estimates for different A/D resolutions were computed, tabulated, and plotted.

RESULTS AND DISCUSSION

Fig. 1 presents the effect of additive Gaussian white noise for deep (left) and superficial (right) fibers. In both cases the bias that affects the estimates obtained by SPM does not depend on the S/N ratio (in the studied range), while the bias increases for decreasing values of S/N when the CC technique is used. It is important to underline that bias is always below 2% of the real value, and thus may be considered negligible for both techniques.

The coefficient of variation increases when the S/N ratio decreases for both SPM and CC, values are always below 8% of the real value, and estimates obtained by means of SPM are always more robust with respect to added Gaussian white noise (paired Wilcoxon test, $p \leq 0.05$).

Fig. 2A presents the effects of CV values on estimation errors. Variation coefficients of the estimates performed by means of SPM and CC are similar and unaffected by the CV values. Percent bias obtained by means of the CC technique is not affected by the spectral shape, while when SPM is used percent bias increases when CV decreases. Again it is important to underline that, although differences are evident, the errors are generally negligible and from a practical point of view the two techniques show a very similar behavior.

Fig. 2B illustrates the effect of quantization errors. It is important to notice that, in the situations considered, no correlation is found between either bias or variation coefficient and the resolution of the A/D converter. This result does not depend on the estimation technique. It follows that even simple and inexpensive acquisition cards (8 bit) can be used to estimate CV without compromising the accuracy of the results.

CONCLUSIONS

The results reported and discussed above show clearly that, in realistic conditions, the two methods show a very similar behavior with respect to additive Gaussian white noise, different CV values, and A/D resolution. The only remarkable difference between SPM and CC is represented by their computational efficiency. Since optimization routines are involved, it is not possible to compare directly the computational cost of the two algorithms, but in terms of CPU time the SPM technique is generally one order of magnitude faster than CC. It is thus concluded that the only significant difference between the two methods is computational efficiency, and that SPM is generally more efficient than CC.

ACKNOWLEDGEMENTS

This work was performed at the Dept. of Electronics of Politecnico di Torino, Italy, in cooperation with the NeuroMuscular Research Center of Boston University, USA. Major support was provided by the Italian Ministry of University and Scientific Research within the framework of the National Project on Rehabilitation Engineering. Partial support was provided by Liberty Mutual Insurance Company.

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MODIFICATION OF THE HUMAN BICEPS BRACHII SPINAL STRETCH REFLEX: EFFECTS ON ANTAGONIST AND SYNERGIST MUSCLES.

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INTRODUCTION

Conditioning of the biceps spinal stretch reflex (SSR) has been examined extensively in normal monkeys by Wolpaw and coworkers (1,2). The biceps SSR can be either uptrained (increased) or downtrained (decreased). In addition, the effect appears somewhat specific to the trained biceps, although its synergists undergo similar but smaller changes (3). Preliminary studies have suggested that normal human subjects can uptrain or downtrain the biceps SSR (4), and that stroke patients with hyperactive biceps SSR's can downtrain the response (5,6). However, neither of the previous human studies examined the effect of training the biceps on synergist and anatagonist muscles. The purpose of the present study is to examine the effects of conditioning the biceps brachii SSR on brachioradialis (synergist) and medial triceps brachii (antagonist) electromyographic (EMG) activity during the biceps SSR interval.

MATERIAL AND METHODS

Ten normal subjects (Table I) met the following criteria: 1) no history of neurologic dysfunction; 2) dominant right upper extremity; 3) no limitation of movement of dominant upper extremity; 4) ability to follow simple commands; 5) no other abnormality affecting movement or movement control; 6) no visual impairment except if corrected by lenses; and 7) no chronic medications that could alter motor function. All subjects signed a consent form approved by the University Human Investigations Committee.

TABLE 1

DEMOGRAPHIC DATA SUMMARY

	Uptraining	Downtraining
Age (years)	25.6 ± 2.1	26.2 ± 3.1
Sex		
Female	2	5
Male	3	0

Subjects were trained to simultaneously maintain the right elbow angle at 90 ± 5° and the biceps integrated EMG at approximately 15% of maximum before the elbow was rapidly extended by a torque motor at random intervals of 1-5 seconds. Each subject had 4 baseline sessions where 250 stretches occurred without feedback of biceps SSR magnitude. Then, subjects were randomly assigned to one of two sequences (Training-Control or Control-Training) and to uptraining (n=5) or downtraining (n=5) groups. The 8 control sessions were exactly the same as baseline sessions. During the 8 training sessions subjects received feedback of the magnitude of the biceps SSR and operant conditioning in the form of rewards for appropriate responses.

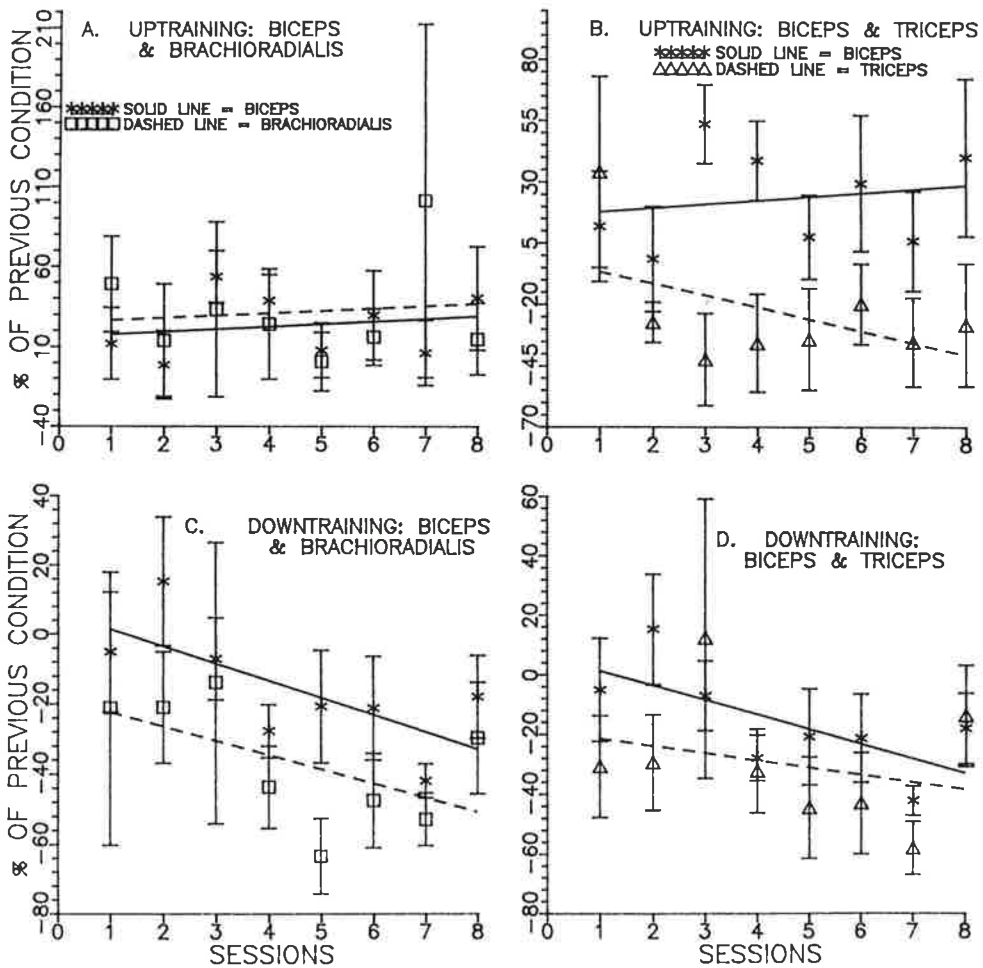

Fig. Graphs of mean percent change (± SEM) from previous condition (baseline or control) for each training session. Each graph plots the biceps versus the brachioradialis or triceps.

RESULTS AND DISCUSSION

Uptrainers increased the biceps SSR by an average of 31% and downtrainers decreased the SSR by an average of 33% from baseline. The average integrated EMG recorded from the brachioradialis during the biceps SSR interval increased during uptraining and decreased during downtraining (Fig). The EMG activity of both the biceps and brachioradialis decreased during 8 control sessions which followed uptraining. However, the reduction in brachioradialis EMG activity was greater. Thus, there appears to be an effect on the synergist (brachioradialis) during training, but the training effect appears more specific to the targeted (biceps) muscle. The antagonist triceps EMG activity decreased during control and training for both uptrainers and downtrainers (Fig).

The present results support the notion that the training effect of operant conditioning is more specific to the trained muscle, but that other synergistic muscles may also be trained. The idea that synergistic muscles can be simultaneously trained while only targeting one muscle may be important in "rehabiltating" tasks versus single muscles after CNS lesions. Future studies will need to examine synergist and antagonist EMG activity during functional tasks and in patient populations.

ACKNOWLEDGEMENTS

This work was partially supported by a grant from the American Paralysis Association.

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EFFECT ON SURFACE EMG WAVE FORMS OF ELECTRODE LOCATION WITH RESPECT TO THE NEUROMUSCULAR JUNCTIONS : ITS SIGNIFICANCE IN EMG-MUSCLE LENGTH RELATION

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Introduction

Spectrum analysis of surface EMG has been used extensively to extract functional and anatomical information from muscle tissue. However, the spectral content may be affected by various factors including muscle length. While Sato, M. (1964)[1], Inbar et al. (1987)[2] and Okada (1987)[3] observed slowing of surface EMG with increase in muscle length in isometric contractions, Sato, H. (1976)[4] did not find any slowing under the similar experimental conditions.

This study was designed to identify and clarify the slowing mechanism of surface EMG wave forms when biceps brachii muscle was lengthened. Using a 128 ch. surface grid electrode and a 16ch. surface electrode array, the location of neuromuscular junctions was estimated and EMG power spectra were computed, under an experimental condition in which the elbow joint angle was varied by 5 steps. As a result, it has been shown that change of surface EMG wave forms when muscle length is altered is due to a change in distance of neuromuscular junctions from the electrodes.

Methods

Three male subjects performed two experiments with a dynamometer composed of a wheel about 60cm in diameter, which allowed the subjects to maintain isometric elbow flexions at various elbow joint angles, with the upper arm held vertical and the forearm supinated.

In the first experiment, the location of neuromuscular junctions in the biceps brachii muscle was estimatied using a 128ch. surface grid electrode, while the subjects maintained isometric elbow flexion of moderate intensity with the elbow joint angle fixed at 90 degrees. This electrode is comprised of a 9$\times$16 grid array of 0.4mm^2 gold-coated metal contacts attached to a cylindrical plate, covering a 40mm$\times$ 40mm muscle surface. The distance between the contacts is 5.08mm in the direction of muscle fibers and 2.54mm in the direction of muscle circumference. Individual EMG signals, 128 in total, were derived bipolarly from a pair of contacts adjacent in the direction of muscle fibers and digitized with sampling frequency of 4kHz and sampling points of 4096 for each channel. The location of neuromuscular junctions were estimated by observing the propagation of action potentials displayed on a graphic terminal (Masuda and Sadoyama, 1988)[5].

In the second experiment, the subjects maintained isometric elbow flexions for 5s at intensities of 25% or 50%MVC, with the elbow joint angle varied by 5 steps between 68 and 158 degrees. EMG was recorded from the biceps with a 16ch. surface electrode array placed longitudinally on the muscle belly after cleansing the skin with alcohol and rubbing electrode jelly. This electrode consists of 17 stainless-steel wires of 1mm diameter arranged in a linear array at intervals of 5mm on a flexible vinyl sheet. Individual EMG signals were derived bipolarly from a pair of adjacent wires, with time constant of 0.03s and cut off frequency of 1kHz, and digitized with sampling frequency of 2kHz. After multiplying individual signals by hamming window, power spectrum and mean frequency were computed. Shift of the location of neuromuscular junctions with changes of the elbow joint angle was detected by observation of the action potential propagation.

Results

Fig. 1 illustrates the location and configuration of neuromuscular junctions of the biceps brachii muscle, as estimated using the grid electrode. The junctions invariably distribute over the middle muscle belly in the transverse direction, but a considerable variation is found in their configurations. The location of neuromuscular junctions in terms of the recording channels, estimated in each subject using the array electrode, is shown in Fig. 2. It is evident that the junctions move to the distal with increase in the elbow joint angle, i.e. lengthening of the muscle, though the process is variable between subjects.

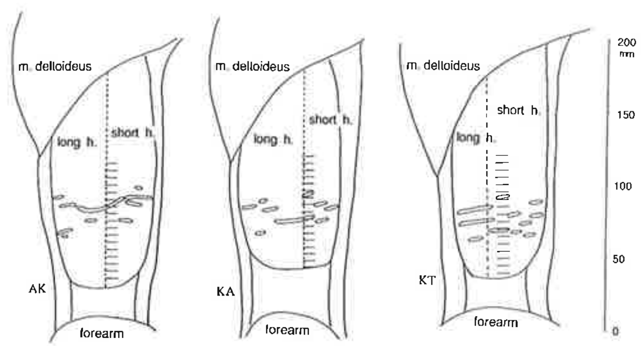

Fig. 1. Distribution and configuration of neuromuscular junctions in the biceps brachii muscle in each subject (AK, KA and KT).

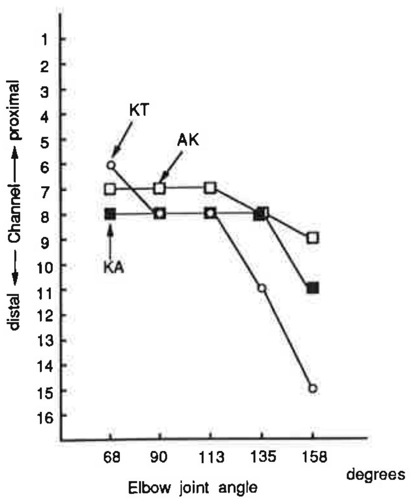

Fig. 2.
Shift of neuromuscular junctions in each subject according to the elbow joint angle.

Fig. 3 shows, for different elbow joint angles, the mean frequency(MF) of EMGs recorded with the 16ch. electrode array from the biceps during 50% MVC contractions in subject AK. With reference to Fig. 2, the MF value is highest in the vicinity of the neuromuscular junctions and reduces drastically with distance from them. Since, as shown above, the junctions move to the distal with increase in muscle length, the channel yielding highest MF value also shifts to the distal. A similar trend was observed in the other two subjects. Behavior of the MF value from each recording channel in response to change in the elbow joint angle is examined in Fig. 4 for the same subject as in Fig. 3. It is clearly shown that MF values of the channels proximal to neuromuscular junctions(1 ~ 6ch.) decline, whereas those of the channels distal to the junctions(9 ~ 15ch.) rise as the joint angle is increased, i.e. muscle is lengthened. Higher MF values are also observed in channels approaching the tendinous portion.

Comparable results with the above were obtained from 25%MVC contractions.

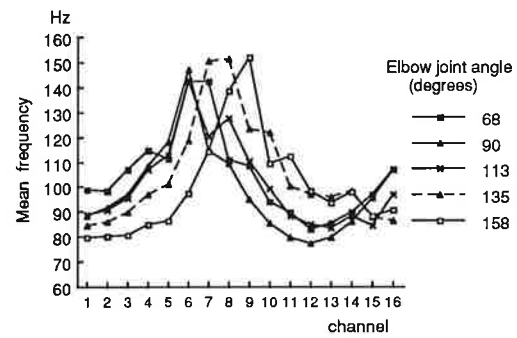

Fig. 3.
The mean frequency of EMG recorded from each channel of the electrode array, as modified with the elbow joint angle :
50%MVC, Subj. AK.

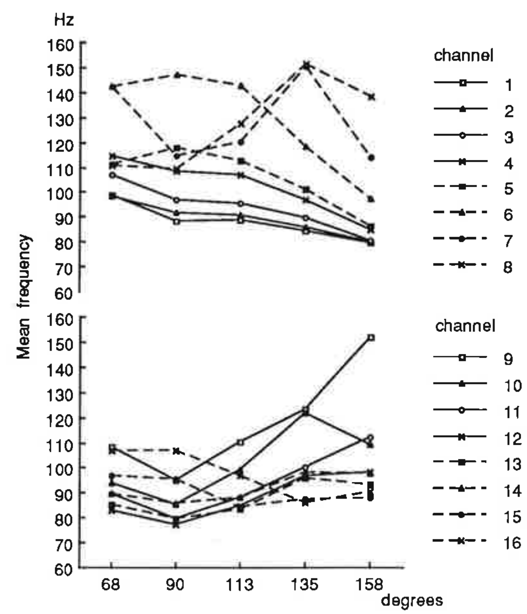

Fig. 4.
Change of the mean frequency of EMG recorded from each channel of the electrode array in response to a change in the elbow joint angle :
Upper figure ; proximal channels.
Lower figure ; distal channels.
50%MVC, Subj. AK.

Discussion

As mentioned in the beginning, the effect of muscle length on surface EMG wave forms has been more or less controversial in the literature. On the other hand, Roy et al.(1986)6 have recently demonstrated that surface EMG wave forms are influenced by the location of electrodes with respect to neuromuscular junctions. These situations warrant a collective analysis of both muscle length and electrode location in terms of neuromuscular junctions.

Our experiments, including analysis of both factors, have revealed that the above two phenomena arise from the same origin. Since the electrode location as regards neuromuscular junctions is altered with a change in muscle length, surface EMG wave forms are inevitably altered. Findings in the litcrature confirming the slowing with increase in muscle length is equivalent to our observation for the channels proximal to neuromuscular juncions. It is presumed that the conventional method places surface electrodes on the skin proximal to the junctions of biceps, thus leading to the slowing with increase in muscle length. On the other hand, the reported failure to observe the slowing conceivably arose from placement of electrodes in the vicinity of the junctions.

Following are possible reasons for the higher frequency near neuromuscular junctions : 1) Action potentials are steeper and shorter in duration near the junctions than at a distance from them. 2) Signals derived from a pair of electrodes crossing the junctions are equivalent to those from electrodes placed closer to each other. 3) When the junctions are placed between a pair of electrodes, differential amplifiers cancel out bidirectionally conducting potentials, yielding signals with a broader frequency content which are sensitive to fast noises. 4) In the vicinity of the junctions, superposition of minutely shifted action potentials gives rise to complex wave forms (Basmajian and DeLuca, 1985)7.

For the lowering of the MF with distance from the junctions, an interpretation offered by Inbar et al.(1987)2 and Okada(1987)3 for the effect of muscle lengthening seems to apply ; the slowing is outcome of an increased MUAP duration caused by a longer distance of propagation which generates a larger time lag between potentials of individual muscle fibers.

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COMPARISON OF EMG POWER SPECTRA WHILE PERFORMING STEPWISE AND RAMP CONTRACTIONS

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INTRODUCTION

In an isometric contraction condition, the influence of force on the EMG power spectrum can be evaluated by 1) comparing power spectra of EMG signals obtained while maintaining a steady level of force (stepwise contraction), this, at different force levels (% of maximal voluntary contraction (MVC)), or 2) comparing power spectra obtained from different portions of EMG signals (corresponding to different % of MVC) produced during a single linearly force varying contraction (ramp contraction). Various results describing the behaviours of both the mean power frequency (MPF) and the median frequency (MF) of the power spectrum across force levels have been found for both types of contractions. In general, it appears that the spectral statistics increase with an increasing level of force in a ramp contraction[1,2]. When subjects are performing step contractions, however, the increase observed is usually either non-existant[3] or less pronounced[4] than that which is observed for ramp contractions.

The goal of the present study was thus to contrast power spectra obtained in these two types of contractions. These contractions were performed within a single session so that the possible effects of other confounding variables (e.g. varying inter-electrode distances (IED), position of the electrodes, muscle length) were minimized.

MATERIAL AND METHODS

Fourteen normal subjects (eigth women and six men) performed a) five ramp elbow extensions ranging from 0 to 100% MVC, each in a 5-s period and b) three 3-s stepwise elbow extensions performed at five different levels of MVC (10, 20, 40, 60 and 80% MVC). These contractions were presented in a random order. To minimize fatigue effects, a 2-min rest period was given to the subject between each contraction.

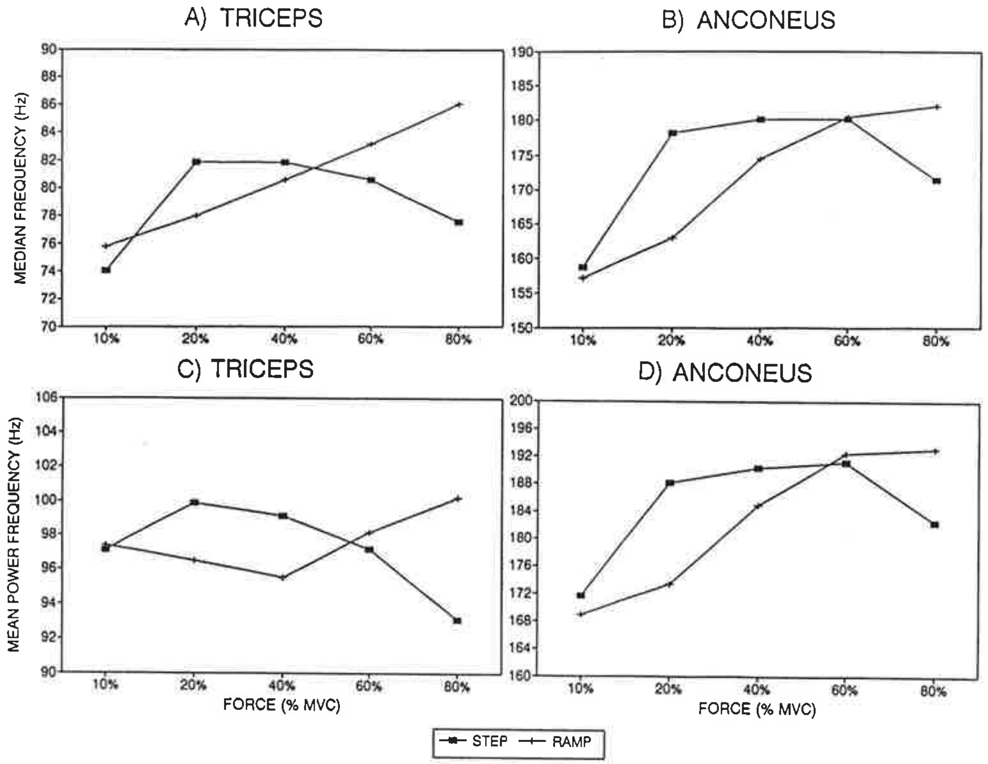

Figure 1. Graph showing the mean values (n=14) of the median frequency (MF) for a) the Triceps brachii (TB) and b) the Anconeus (AN) and the mean power frequency (MPF) for c) the TB and d) the AN, with an increasing level of force, for the two types of contraction (RAMP/STEP). The score for a given subject is a mean of five consecutive contractions for the ramp condition and three consecutive contractions for the step condition. The coefficients of variation (CVs) (SD/X*100) reached 15% on average.

EMG signals from the Triceps brachii (TB) and the Anconeus (AN) muscles were picked up with surface miniature electrodes placed longitudinally to the muscle fibers, 6 mm appart (centre-to-centre). The ground electrode was placed over the right lateral epicondyle. The TB and the AN EMG signals were pre-amplified and amplified with a frequency band ranging from 16 to 800 Hz (TECA Mark III unit, CMRR = 90 dB) and digitized on-line with a sampling frequency of 2000 Hz (PDP11/23+ computer).

For both types of contraction, spectral analysis (Hamming window processing, 512 points, fast Fourier transform) of the EMG signals

of the TB and the AN was performed on 256ms windows taken at 10, 20, 40, 60 and 80% MVC. The MPF and the MF of the power spectrum were calculated for each contraction.

RESULTS

Two-way ANOVAs for repeated measures performed on the ramp versus step data did not depict any significant differences (p>0.05) in the values of either the MPF or the MF between the two types of contractions, this for the three levels of force tested (10, 40, 80% MVC). In contrast, these ANOVAs depicted significant increases (p<0.05) across force levels for both the MPF and the MF values of both muscles for both types of contractions. An exception was found for the MPF of the TB data (p>0.05). Even though there were no overall significant differences for the MPF and the MF values between the ramp and the step contractions (p>0.05), significant interactions (p<0.05) were found between the two factors, that is, type of contraction and level of force. These interactions point out the existence of different behaviours for both the MPF and the MF across levels of force, this according to the type of contraction being performed (Fig.1).

DISCUSSION & CONCLUSION

Certain factors such as IED[1], varying thickness of the skin over the different muscles investigated[1] and others, can explain the different results observed among studies concerning the variability of the behaviour of the MPF and the MF with an increasing level of force. From the present results, it is obvious that the type of contraction being performed (ramp versus step) has a major influence on this behaviour. The spectral statistics were, as previously reported[4], observed to a) increase rapidly at very low force levels (10% to 20% MVC) and then b) level off and even decrease at the higher levels of force during step contractions. This in contrast to the progressive increase up to levels approaching 80% MVC observed for the ramp contractions, as previously reported[2].

Consequently, it can be speculated that the observation of a generally lesser increase of the spectral parameters with step contractions as compared to that observed with rámp contractions could partly be explained by the type of contraction performed. Since the increase of the spectral statistics has been suggested to reflect the gradual recruitment of larger diameter muscle fibers[2,4,5], the different patterns of increase observed between the two types of contraction may reflect different motor unit recruitment strategies. Furthermore, since the increases of the MPF and the MF of the AN are generally more pronounced than for the TB (see Fig.1), it can be suggested that recruitment plays a more important role as a force generating mechanism[5] for this muscle than for the TB, which would rely more on rate coding. Some observations on the firing rate of both muscles have been made to support this[6].

ACKNOWLEDGEMENTS

This work was supported by the Quebec March of Dimes and Fonds de la recherche en santé du Québec (FRSQ).

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Electromyographical Kinesiology
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APPLICATION OF NEURAL NETWORK MODELS TO EMG SIGNAL ANALYSIS

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INTRODUCTION

Neural Network models are commonly used as pattern classifiers for extracting and classifying features. The ability of these models to capture non-linear mappings and generate spontaneously useful global computational functions such as classification, optimization, and control has been documented in the neural network literature (1,2,3). The computational properties of neural networks have been successfully applied to applications such as sensory/motor control (4,5) and (adaptive) pattern recognition (2,5). In addition, kinematic (6), dynamic (7,8), and speech recognition problem solving (9) with the use of the neural network model of backpropagation (BEP) have recently been documented. The successful application utilizing backpropagation, a supervised neural network, for modeling the EMG to joint torque relation in the human ankle joint is presented in this paper.

BACKGROUND

We address the feasibility of designing a "neural network" which will obviate the need to specify the EMG-force relationship apriori. There are disputes in the literature of the dependency of muscle force on processed EMG (10,11,12,13). Some approaches suggest a linear relationship, while others have reported relations of higher order. The employment of EMG signatures for controlling functional electrical stimulation (FES) in paraplegics in a manner that recognizes and executes the patient's intended limb functions and compensates for muscle fatigue is currently being investigated (14,15,16,17). Myoelectric prosthetic limbs have been developed which utilize spatial patterns of signals from muscles which remain after a limb amputation (18). Some of these approaches include Pattern Recognition (19), Model Reference (20), and Autoregressive Models (21).

The promising features of a neural based approach to adaptive control of myoelectric prosthesis, orthotic devices or robotic manipulators include: 1. control laws that are not explicitly stated since learning occurs by showing examples; 2. fault tolerance provided by massive parallelism; 3. robustness to unmodeled parameters due to the network's generalization properties; and 4. abundance of local minima in the network state space used for content addressability and retrieval of information under noise and uncertainty (8,22).

METHODS AND INSTRUMENTATION

An experiment was designed to measure the ankle EMG-joint torque relations at the full range of ankle movement and full range of torque output under supine condition.Three subjects were tested. Measurements from six muscle sites (tibialis anterior, peroneous longus, flexor digitorum longus, gastrocnemius-medial, gastrocnemius-lateral, and soleus) about the subjects ankle joint and the torque were recorded with the use of KinCom II dynamometer and MicroVax II minicomputer. The data presented is with the ankle angle set at 0 degree. The EMG signals were full wave rectified (Instech EMG amplifier system), bandpass 100-1000 Hz, utilizing surface EMG electrodes. Joint torque, angle, and EMG about the ankle were sampled at 100 Hz.

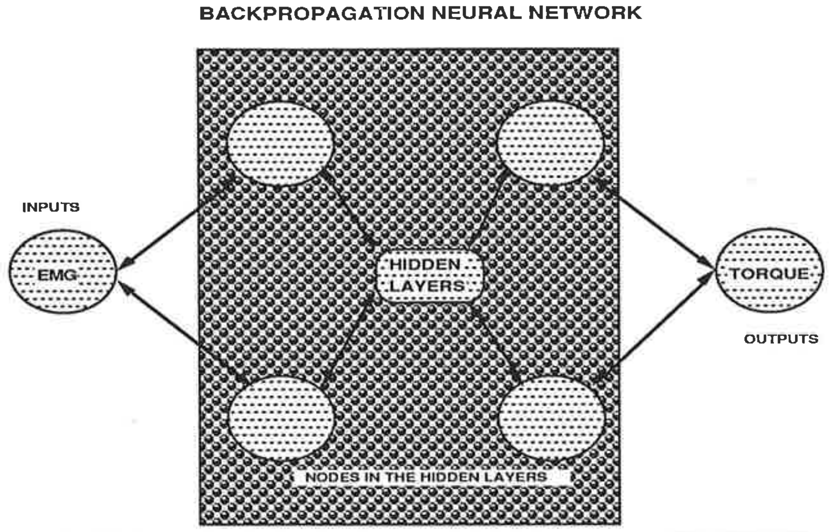

Figure 1. BEP General Architecture for Biomechanical Model of EMG-TORQUE

RESULTS

The backpropagation neural network model simulated a multilayer perceptron for solution to the EMG-torque mapping of the ankle joint under supine condition.The architecture of the network is 6 inputs (muscle signals),2 hidden layers (6 nodes/layer),and 1 output (torque) (Figure 1). One subject's data is presented.

Trials of dorsiflexion and plantarflexion signals were windowed and 100 samples/trial (representing 1 sec of data) were extracted for the training signal. The training signal consisted of 4000 presentations of many input (EMG) - output (torque) examples (samples) from +1000 N-m through -1200 N-m, randomized with respect to a normalized torque. This supervised

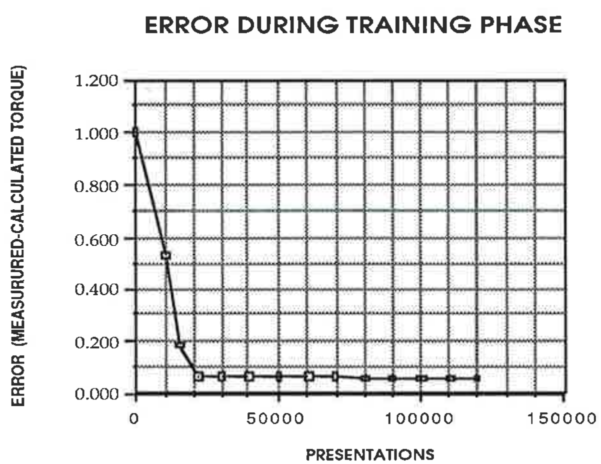

Figure 2 represents the error during the training phase of the backpropagation model.

neural network model is complete when the weight changes approximate 0 and the change in the predicted torque from the experimentally measured torque approximates 0 (at 120,000 presentations, error stabilizes at 6%). A more finer prediction during the training phase is achieved by decrementing the momentum and learning rate parameters (initially set to 0.4 and 0.8), once a rapid decrease in the error results, i.e., at the beginning of learning (see Figure 2, at approximately presentation 16,000).

Embedded within the trained network is the mapping of the muscle-torque relation. These mappings may be considered as the library of learned solutions of the EMG-torque relations of the ankle joint. The muscle signals (input) may be considered the intent of the system while joint torque (predicted output) is the "controlled" variable (Figure 3).

These results detail the use of backpropagation as a model for prediction of both direction and level of joint torque from EMG about the ankle joint. The use of BEP for synergy classification has previously been presented and has been suggested as a tool for classification of EMG synergies for clinical database applications (23).

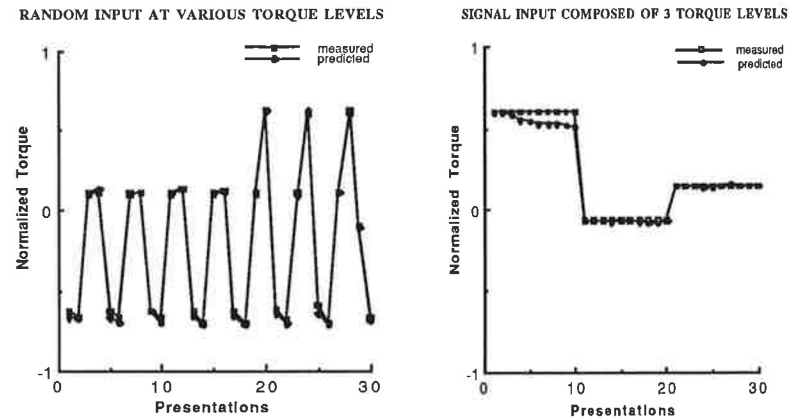

Figure 3 presents the measured and predicted torque response from the EMG data presented feedforward to the trained network from samples that are random and at distinct torque levels.

CONCLUSION

The successful implementation of an EMG-torque model with BEP, demonstrates fast system response/convergence, and minimal over/under shoot (when the model has been successfully trained) (see Figure 3). This may promote more accurate and immediate torque response from orthotic/prosthetic controllers, which is supported by adaptive robotic control applications utilizing BEP (22). EMG modeling issues such as coactivation and multijoint muscles are not problematic within a parallel processing model such as BEP. With the employment of a heirarchy/feedback BEP model, robustness in the controller may be feasible. Tuning of an EMG controller artificial device, to correct realtime factors due to muscle fatigue, sweat, gain changes, and EMG drift (10,11,14,15,16,17,20), may prove to be highly successful with this modeling approach.

The application of neural learning algorithms in the controller for intent recognition systems such as robotic manipulators, myoelectric prostheses, and functional electrical stimulators may be appropriate. In fact, for applications such as these, it is desirable to have an adaptive controller that learns to control the system while in operation. This may be achieved through heirarchy of control and through variations of the learning procedure of the neural net backpropagation model which desires state information rather than error gradients (22,24,25).

ACKNOWLEDGEMENTS

Research partially supported by Grant AFOSR #890010 and Calhoun Research Endowment. We gratefully acknowledge the use of laboratory facilities and technical expertise at the FNS Laboratory at Shriners Hospital, Philadelphia Unit, and the use of laboratory facilities at the Gait and Motion Analysis Laboratory, Moss Rehabilitation Hospital, Philadelphia.

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NORMALIZED ELECTROMYOGRAPHIC ACTIVITY PATTERNS IN HUMAN EXTENSOR CARPI RADIALIS LONGUS AND FLEXOR CARPI RADIALIS MUSCLES: DIFFERENTIAL ACTIVITY

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INTRODUCTION

The concept of compartmentalization stems from observation that cat muscles are organized into discrete morphological partitions. Each partition is supplied by a single primary branch of a muscle's motor nerve, and contains a unique population of motor unit types[1]. Electromyographic (EMG) studies of cat lateral gastrocnemius muscle found differential activity of individual compartments during gait[2] and postural disturbances[3].

Our first effort at a human model of neuromuscular compartmentalization is based upon previous cadaver studies[4] which demonstrated a consistent proximal/distal branching pattern of the primary nerve branches to human Flexor Carpi Radialis and Extensor Carpi Radialis Longus. We related these branching patterns to normalized EMG activity from the muscle partitions surrounding the primary branches within the ECRL and the FCR.

METHOD

In situ dissections of 16 ECRL and 9 FCR human cadaver specimens revealed a proximal/distal innervation pattern, consistent with those of previous investigations[4]. Ratios relating the muscle belly length to the bony forearm length (FCR=0.58; ECRL=0.38) were calculated from cadaver measurements. Two bipolar fine wire electrodes (25 um uninsulated, 1 mm tip) were inserted in the center of the proximal/distal muscle territories, using the above ratio as a constant for each muscle and dividing the estimated muscle belly length in half.

Twenty subjects (9 males, 11 females) performed three repetitions of twelve randomized forearm, wrist, and finger movements. Recorded raw EMG data were full-wave rectified, integrated at 20 ms intervals, and averaged. Integrated EMG data were normalized as percent of activity produced by a maximal isometric contraction for each muscle. Means of normalized EMG data for the three repetitions of each movement were computed. These means were compared using a two-way analysis of variance (ANOVA). Post-hoc Tukey's tests were then calculated.

To determine if true differences in motor unit activity occurred between partitions, spike triggered averaging (STA) was performed on the EMG data of three additional subjects (2 males, 1 female). These subjects attempted to iso-

TABLE 1
Mean Normalized EMG Values for ECRL - Proximal (pECRL) and Distal (dECRL)(n=20)

Movements	pECRL $\bar{X}$	SD	dECRL $\bar{X}$	SD	F
Elbow flexion	5.10	4.24	3.34	2.82	5.30*
Elbow extension	4.25	3.53	2.42	1.85	4.91*
Wrist flexion supported	5.62	4.01	4.34	3.82	3.60
Wrist flexion unsupported	5.58	3.48	4.21	3.60	4.62
Wrist extension supported	9.09	4.97	6.94	3.88	8.77*
Wrist extension unsupported	7.93	4.36	7.25	5.54	6.76
Radial deviation supported	5.38	2.83	3.21	1.38	8.14*
Radial deviation unsupported	5.94	3.34	4.22	2.37	6.90*
Ulnar deviation supported	3.87	2.85	1.96	1.26	7.72*
Ulnar deviation unsupported	4.50	3.10	2.75	1.80	5.06*
Finger grasp supported	7.50	4.45	8.51	7.50	2.03
Finger grasp unsupported	10.69	6.16	10.51	7.41	3.31

*$p < .05$

TABLE 2
Mean Normalized EMG Values for FCR - Proximal (pFCR) and Distal (dFCR) (n=20)

Movements	pFCR $\bar{X}$	SD	dFCR $\bar{X}$	SD	F
Elbow flexion	4.48	4.13	2.94	2.93	5.30
Elbow extension	3.78	2.69	2.68	2.44	4.91
Wrist flexion supported	7.03	3.69	6.04	3.71	3.60
Wrist flexion unsupported	6.81	3.63	5.25	2.63	4.62
Wrist extension supported	5.68	3.95	4.18	2.95	8.77
Wrist extension unsupported	5.86	4.84	3.77	2.62	6.76
Radial deviation supported	4.56	3.37	2.97	1.99	8.14*
Radial deviation unsupported	4.63	3.44	3.10	2.13	6.90
Ulnar deviation supported	3.86	3.12	2.34	1.95	7.72*
Ulnar deviation unsupported	3.64	3.51	2.49	1.89	5.06
Finger grasp supported	7.07	5.62	5.55	5.22	2.03
Finger grasp unsupported	7.47	6.26	6.98	7.45	3.31

*$p < .05$

late single motor unit activity within each partition using single channel audio and visual feedback.

RESULTS

Significant ($p < .05$) differences in normalized EMG activity between proximal and distal sites in the ECRL (Table 1) were found for most movements. Although significant differences between partitions in the FCR (Table 2) were found only in wrist radial and ulnar deviation movements, the proximal partition showed consistently greater activity then the distal partition.

Spike triggered averaging results demonstrated non-time-locked single motor unit activity (Fig. 1) in separate partitions of both the ECRL and FCR. Oscilloscope observations of these data revealed specific increments and decrements in raw EMG within proximal and distal partitions during certain wrist and finger movements.

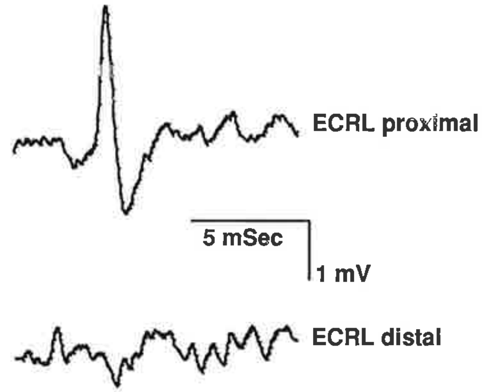

Fig. 1. Spike triggered averaging data of 1 subject isolating single motor unit activity in the proximal ECRL. Trigger is pECRL average is dECRL.

DISCUSSION

Partitioning of human forearm musculature may have a functional significance. In most movements, the proximal partitions of both the ECRL and FCR muscles showed greater activity than distal partitions. Electromyographic data normalization should have controlled for the contaminating effect of differences in muscle girths between partitions. However, during unsupported wrist extension and finger grasp, normalized activity of the proximal and distal ECRL equalized indicating a change in activity with different tasks. Seyfarth[5] obtained similar findings in the FCR, biceps brachii, and flexor carpi ulnaris muscles. Results from STA demonstrate that individual motor units in either proximal or distal partitions may be selectively recruited. During this study subjects reported having to change the orientation of the wrist in order to selectively recruit single motor units in partitions of the same muscle, suggesting a kinematic diversity of muscle partitions.

One may explain the differential activity between muscle partitions by the fiber architecture of the individual muscles. If separate partitions of a muscle have varying angles of pennation, the effective force on the tendon should be different for each partition. Muscle partitioning based on innervation patterns may be one method for the central nervous system (CNS) to vary the individual tendon forces in order to refine movement. If one adds the concept of muscle partitioning to existing literature regarding coordinated movement, muscle function can be viewed as more than the orderly recruitment of motor units. Partitions of an individual muscle may be differentially recruited in order to produce complex or discrete movements or postures.

Future research into muscle partitioning may need to focus on motor unit types of partitions, replication of these results in various human muscles, and additional functional implications of these partitions. If human muscle partitioning exists, then rehabilitation techniques may need to be re-examined with respect to electrical stimulation, biofeedback, and other muscle reeducation techniques.

CONCLUSIONS

Separate partitions of the same muscle can be differentially activated during the same movement, supporting the possibility that neuromuscular compartments exist in human forearm musculature. These muscle partitions may have distinct functional roles.

ACKNOWLEDGEMENTS

A special thanks to James Hudson for his technical assistance on this project and Lisa Walker for her help with the manuscript.

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CLINICAL ELECTROMYOGRAPHY

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

KNOWLEDGE-BASED SYSTEMS FOR EMG.

NEUROANATOMY KNOWLEDGE-BASE: A COMMON GOAL FOR CLINICAL AND KINESIOLOGICAL USE

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INTRODUCTION

Computer-based medical systems are used on-line in kinesiology and clinically for keeping track of interventions and responses from the human being. The external stimuli may be visual, auditory, tactile or electrical stimuli to the patient. Methods for quantification of the responses from muscle and nerve have a wide range of applications for diagnosis and for the study of motor control and performance.

Intelligent computer-based medical systems or knowledge-based systems have all the functionalities described for the computer-based medical systems. In addition, they have inference capabilities and may use the knowledge from several knowledge bases to reason about problems and to provide the researcher or clinician with decision support (Fig. 1).

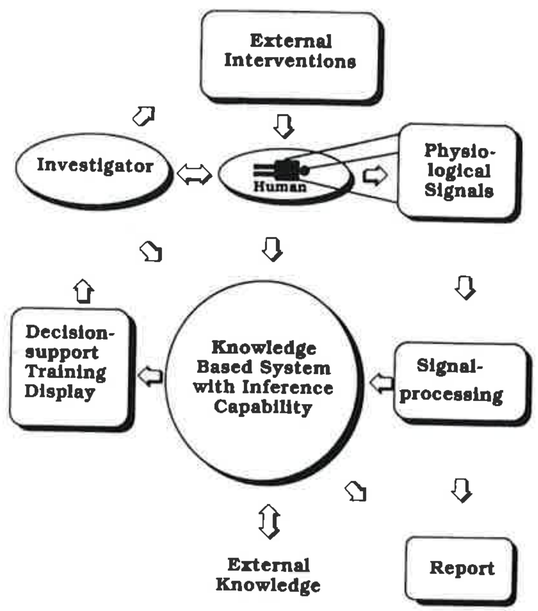

Fig.1. Intelligent computer-based medical system.

MATERIALS AND METHODS

Most prototype knowledge-based systems, developed for clinical neurophysiology, have been rule-based systems which simulate human expert reasoning. It was experienced, that they were not effective in capturing causal knowledge, as it is usually represented in textbooks and presumably in the minds of clinicians.

The MUNIN system is based on causal probabilistic networks (2). In the network, the nodes represent disorders, pathophysiology and test results. The network is called causal because the links between the nodes indicate causal connections in a broad sense, and it is called probabilistic because the causal connections are expressed as conditional probabilities. The probabilities of all nodes are calculated by Bayesian probabilistic methods, using the HUGIN shell (1).

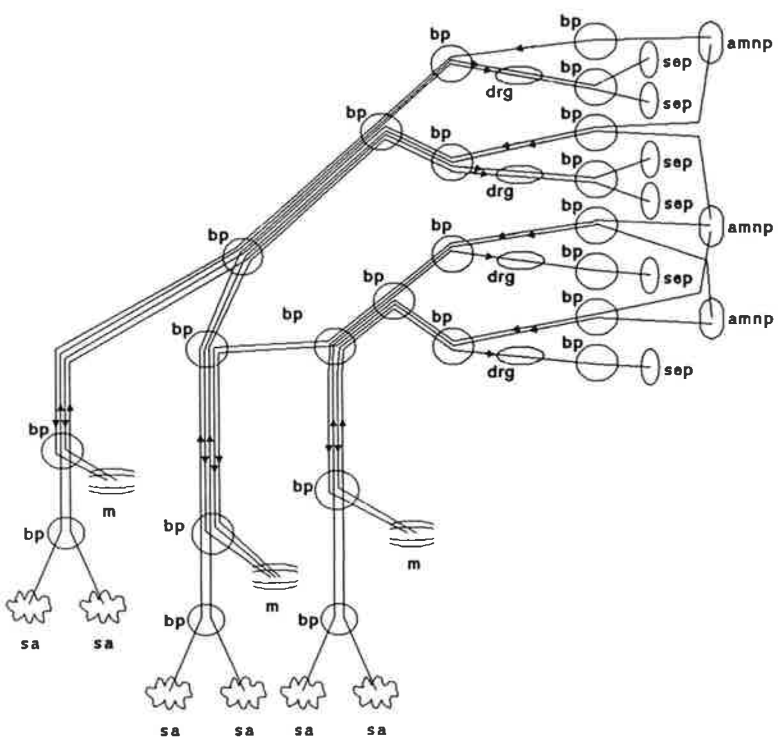

Fig. 2. Schematic drawing of the computerized neuroanatomical network consisting of *nodes* and *links*. There are six different types of *nodes* and two different types of *links*.(from Woldbye et al. 1989).

To develop a knowledge-based system for clinical neurophysiology, it was necessary to formalize the key concepts in the diagnostic process: diseases, pathophysiology and test results. The performance of the diagnostic network was tested on cases collected by seven experienced clinical neurophysiologists from six European countries. Out of 70 cases collected by the experts, eleven were chosen for a peer review, where the diagnostic performance of MUNIN was compared to the diagnostic performance of the experts. The MUNIN system performed as well as the neurophysiologists and they stated that the "input of the cases is considered to be sufficiently representative for a judgement to be made" and that they "consider that MUNIN performed at the same level as an experienced neurophysiologist". MUNIN dealt particularly well with patients with multiple disorders.

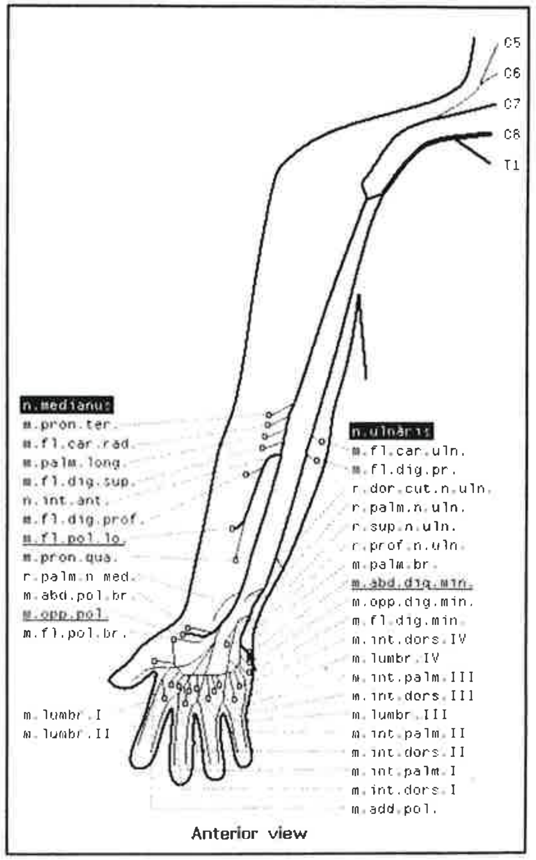

Fig. 3. Tracing of paths from structures to the spinal cord. By pointing at m. flexor pollicis longus, at m. opponens pollicis and at m. abductor digiti minimi, the efferent paths from the alpha motor neuron pools to the three muscles are highlighted. The paths shared by the three efferent paths are emphasized. (from Woldbye et al. 1989).

We suggest the development of a knowledge-based system for kinesiology. We expect that concepts might be similar and that some knowledge bases may be common. In particular the anatomical knowledge-base designed for clinical EMG (4) may be shared with kinesiology and rehabilitation. We have tried to make it as complete and accurate as possible (Fig. 2). It includes 285 muscles and 120 sensory areas, and all nerves innervating these structures. Anatomical anomalies that have been described in the literature, are also included. At present, the system does not include the peripheral autonomic nervous system.

The system can be implemented on a PC-based computer with high resolution graphics. In clinical neurophysiology it can be used for guidance. When an abnormal finding is recorded in one or more muscles, the user may want to trace the nerve fibres innervating these muscles (Fig. 3). It can be mouse-operated and may include cross-sectional pictures of the limbs, essential considering the placement of electrodes and when artefacts due to volume conduction in tissue may be of importance for interpretation of kinesiological data (3).

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

EVALUATION OF SEGMENTAL MOTOR BLOCKADE DURING EPIDURAL ANESTHESIA BY MEANS OF QUANTITATIVE ELECTROMYOGRAPHY

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INTRODUCTION

Epidural anesthesia, or analgesia, is a technique that is commonly used for certain surgical procedures. Epidural anesthesia offers many advantages over general anesthesia. A major concern regarding epidural anesthesia is the degree of blockade of the sensory as well as the motor nerve trunks. The necessity of appropriate sensory blockade, for surgical procedures, is obvious. Good blockade of the motor nerves is a prerequisite for smooth surgery. This study has concentrated on the blockade of the motor nerves.

Axelsson, Hallgren, Widman, Johansson and Olstrin (1) at the Örebro Medical Center have earlier described a method for quantification of the motor blockade of muscles innervated from the L1 to S2 spinal levels. Dr. Axelsson's method was based on maximal isometric muscle force measurements for hip flexion, knee extension and plantar toe flexion. Prior to epidural injection a reference level was determined for the movements studied. With this technique it was possible to produce maps over the temporal and spatial motor blockade behavior of different anesthetic agents for the L1 to S2 spinal levels. In this paper a technique to monitor the motor blockade at higher levels, based on EMG from the rectus abdominis muscles, is presented. A similar technique, applied on bupivacaine, has been demonstrated by Chamberlain and Crawford (2).

MATERIAL AND METHODS

In this study a new local anesthetic agent, ropivacaine, was evaluated in 30 volunteers who had given their informed consent. 20 ml of three different concentrations of this drug, 0,5%, 0.75% and 1.0% were administered epidurally at the L2-L3 interspace. There were ten volunteers in each group.

It is well known how the average rectified EMG relates to exerted muscle force when force is voluntarily moderated. It is not obvious how EMG and force are related when the exerted force is reduced by an epidural nerve blockade. In order to study this relationship the surface electrode detected EMG and knee extension force were recorded simultaneously during maximal voluntary contraction (MVC) of the quadriceps muscle while the subject received epidural anesthesia.

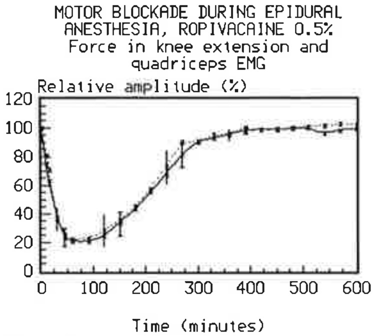

Fig. 1

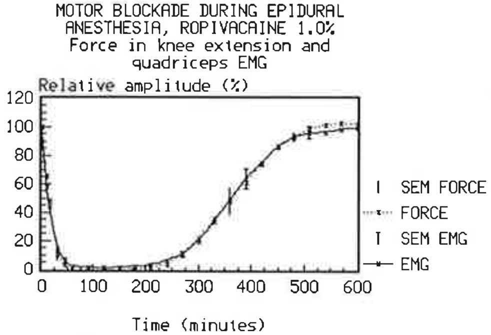

Fig. 2

Figures 1 and 2. Time courses during epidural anesthesia for knee extension force and average rectified EMG in the quadriceps muscle for 0.5% and 1.0% ropivacaine. The ordinate is in percent of the reference level, the abscissa is in minutes after epidural injection.

Figures 3, 4 and 5 shows the development of the abdominal and quadriceps EMG during epidural anesthesia for the three groups investigated. As shown in these figures the blockade close to the injection level (L2-L3) is more profound than at the more distant levels.

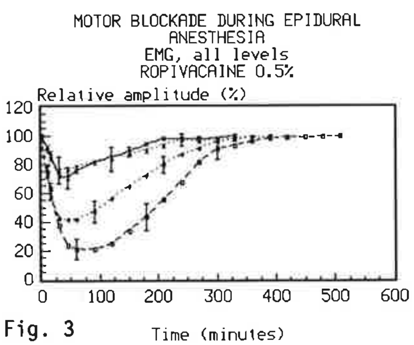

Fig. 3

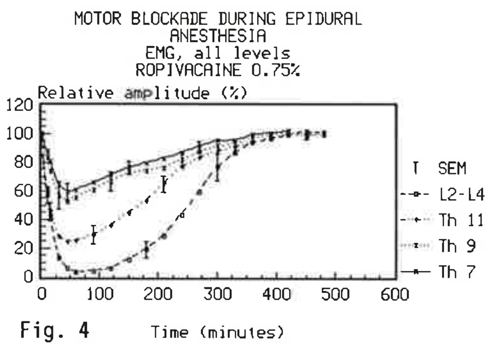

Fig. 4

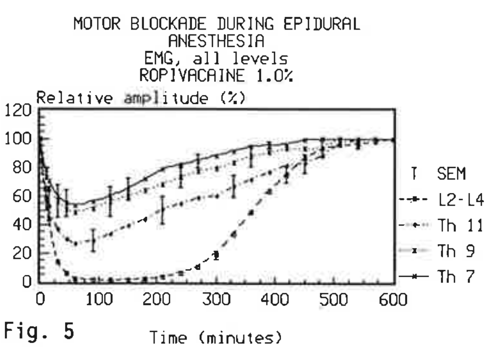

Fig. 5

Figures 3, 4 and 5. Average rectified EMG from rectus abdominis and quadriceps during epidural anesthesia, ropivacaine 0.5%, 0.75% and 1.0% respectively.

The amplified EMG signal was sampled at 3750 Hz per channel into a digital computer where the average rectified value was calculated. The acquisition was done in three 1 second chunks during the MVC, the mean value of the average rectified EMG for the three seconds was used.

Based on the assumption that the average rectified EMG was a good measure of the exerted force in the quadriceps muscle it was anticipated that the motor blockade of the rectus abdominis muscles could be measured using a technique analog to the force technique used for the lower limbs. In order to deal with the higher spinal levels, surface electrodes were applied bilaterally on the abdomen. The procedure is described in detail by Nydahl, Axelsson, Hallgren, Larsson, Leissner and Philipson (3). In this way six channels of EMG was recorded from the abdomen. The mean value of the left and right observations in each subject was used.

The MVC levels for the rectus abdominis muscles were determined in analogy with the force measurements in (1), this was achieved by asking the subject to perform a maximal isometric situp against a strap over the chest. EMG was recorded while the subject performed the situp. The MVC was repeated 4 times before the epidural injection, the mean EMG values from these four repetitions were used as the reference levels. After the epidural injection, EMG was recorded during MVC at regular time intervals for the duration of anesthesia. In this way it was possible to produce motor blockade maps from the abdominal level. With the addition of quadriceps EMG a picture of the motor blockade at 4 different levels was achieved.

RESULTS

The time courses during epidural anesthesia for knee extension force and average rectified EMG in the quadriceps muscle for 0.5% and 1.0% ropivacaine are shown in Figures 1 and 2. Each figure represent a group of ten volunteers. As shown in the figures the average rectified EMG seem to be an accurate representative of the exerted force in the epidurally partially blocked muscle.

By adding the forces from toe and hip flexion a rather complete picture of the motor blockade at different levels is achieved, this is examplified in Figure 6 for 1.0% ropivacaine.

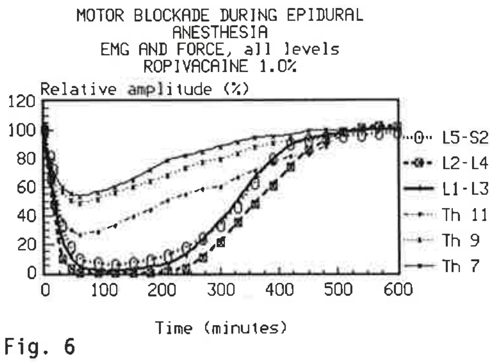

Fig. 6

Figure 6. Motor blockade in ten subjects for the S2-Th7 spinal levels, 1.0% ropivacaine.

CONCLUSION

It seems that the described method for quantification of motor blockade at the abdominal level is a technique with great potential. With this technique it has proven possible to produce maps over the behavior of different drugs in the motor block sense. These maps may guide the anesthesiologist in the selection of the optimal epidural drug, or concentration, for any particular surgical procedure.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

ELECTRICAL ACTIVITY IN DENERVATED MUSCLE

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INTRODUCTION

Extracellularly recorded spontaneous electrical activity has seldom been studied in artificially denervated muscles of animals. Here we present quantitative data of fibrillation potentials aimed to examine whether the spontaneous activity belongs to single fibres and to derive quantitative parameters during denervation.

Modelling the extracellular representation of muscle fibre activity (1) important aspects are e.g.: the intracellular action potential (the extracellular action potential is globally linear with the second derivative of the intracellular action potential) and the cross sectional area of the muscle fibre. The experimentally found changes are for the intracellular action potential a decrease in amplitude and less steep slopes (2, 3) with a simultaneous shift of the resting membrane potential, to less negative values (in the majority of papers, e.g. 2 and 3), and for the mean cross sectional area of the muscle fibre a decrease (3). As a result apparent changes in the extracellular action potentials due to denervation have to be expected.

In our study spontaneous activity and evoked single fibre action potentials (SFAP's) have been recorded in the same muscle. The SFAP's have been evoked with intracellular current injection. The stimulating micropipettes have been used for recording purposes as well. A few histochemical data will be given.

MATERIAL AND METHODS

Denervation. Denervation was introduced 7 or 10 days prior to the experiment in the right hind limb in 9 and 4 muscles resp. About 5mm. of the n. peroneus was removed; the proximal stump of the nerve was sutured to muscular tissue.

Muscle preparation. All experiments were done with the m. extensor digitorum longus (EDL) in the hind limb of the rat (Wistar, adult males). The rats were anaesthetized intraperitoneally with pentobarbital sodium. The muscle was *in vivo* as described before (4). The muscle temperature was 36–38°C. The length of the muscle was a few mm. above optimum twitch length.

Micropipette electrode. The tip diameter of the micropipette electrodes was about 2 μm. The pipette was filled with 0.5 M KCl. Activity of a penetrated fibre was evoked with a hyperpolarizing current pulse (amplitudes ranged from 70 to 300 nA).

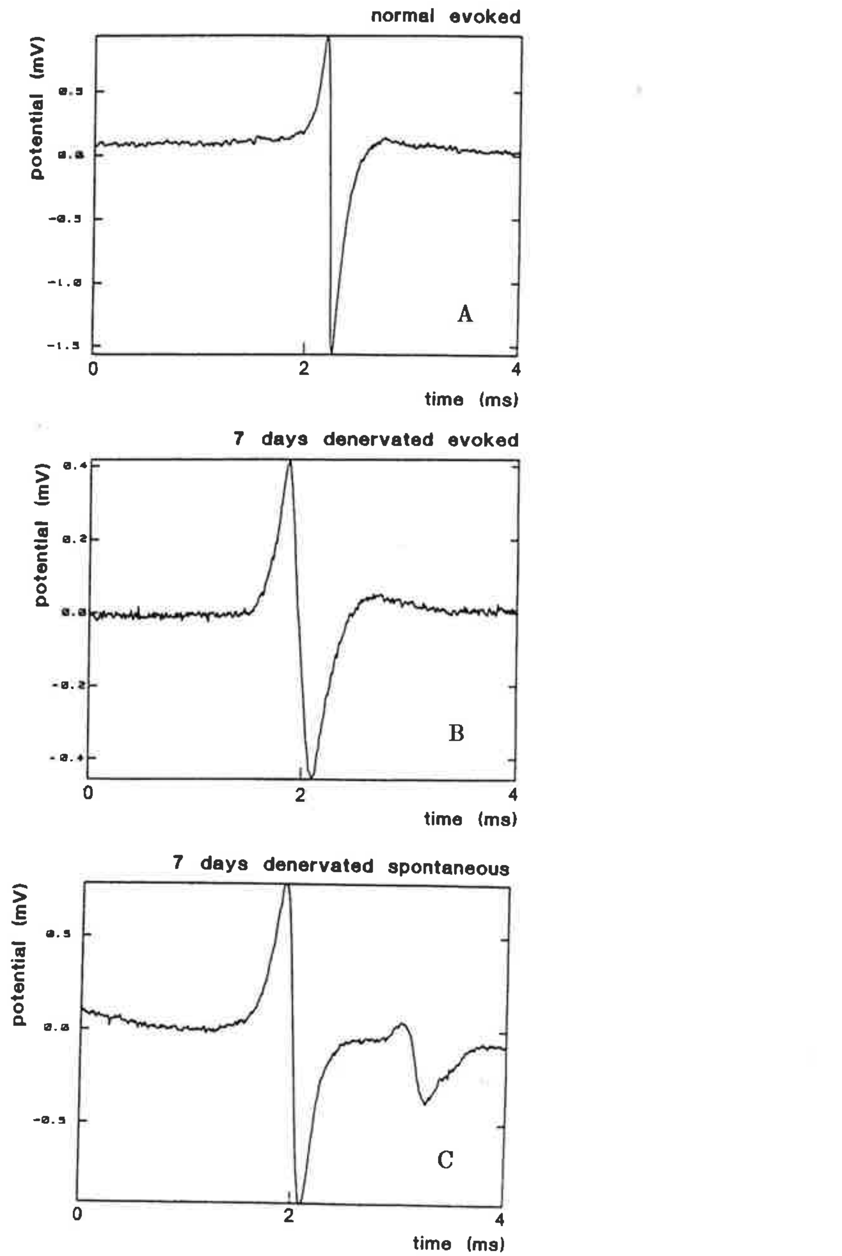
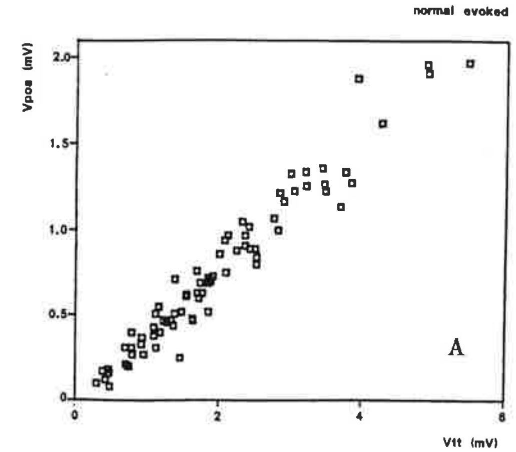
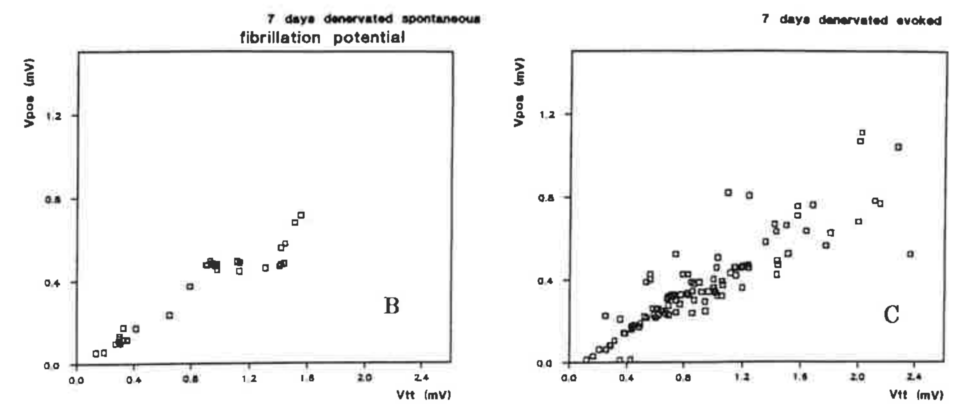


Fig. 1. Examples of extracellular recordings with wire electrodes. A and B show SFAP's, evoked with intracellular current injection. C shows spontaneous activity: fibrillation potentials.

Fig. 2. Plots of Vpos versus Vtt for normal and 7 days denervated muscle. Note that that the scales in A differ from those in B and C. A and C show data of SFAP's, evoked with intracellular current injection. B shows data of spontaneous fibrillation potentials.





Wire electrodes. The extracellular recordings were made with stainless steel wires with a diameter of 25 μm, isolated with 4 μm Karma coating. The tip was cut obliquely. Up to fourteen wire electrodes were inserted in the muscle.

Signal analysis. All recordings were sampled with 100 kHz. The characteristics derived were: the resting membrane potential prior to stimulation from the intracellular pipet signal and the top–top amplitude (Vtt), the amplitude of the first positive phase (Vpos), and the time between the first positive and the negative peak (Δt) from the extracellular wire recording.

Morphologic data. The morphology was studied in 10 μm thick, Sirius Red stained from the cryo–sections. An estimation of the mean diameter was made for 100 fibres in the center of the muscle. The extracellular fraction was determined in the same area. Only one muscle was used per 7 and 10 days denervation, their contralateral muscles were used as control.

RESULTS

Denervation check. In the denervated hind limb toe spreading and extension was absent. It was not possible to evoke a muscle contraction by stimulating of the nerve.

Resting membrane potentiale. The mean resting membrane potential of 7 days denervated muscles was 9 mV higher than the normal value (mean –74 mV, number of fibres 50). We found no difference between the mean values of 7 and 10 days denervation (both –65 mV, number of fibres 424 and 53 resp.).

Comparison fibrillation potentials and SFAP's. In normal muscle we recorded evoked SFAP's (fig. 1A) and only very seldom spontaneous activity. Fig. 1B shows an evoked SFAP for denervated muscle. In all denervated muscles fibrillation potentials were found (fig. 1C). Sharp positive waves were rarely seen.

Some characteristics of the fibrillation potentials and evoked SFAP's were compared. The Vpos and Vtt data have been plotted in fig.2. All parts of the figure show a linear relation between the two variables, the slopes of the lines are about equal. Note that the figures 2B and 2C for 7 days denervation do not differ and the same remark holds when comparing the evoked and spontaneous potentials of 10 days denervated muscles (not shown).

With respect to the Δt versus Vtt plots we found no clear differences between the fibrillation potentials and the evoked SFAP's (not shown). In general the value of the Δt is smaller for higher Vtt. In order to compare the Δt of the three experimental sets the mean Δt was determined for all Vtt's $\lesssim$ 1.6 mV. The mean value of Δt changed remarkably due to 7 days denervation, from 120 μs (± s.d. 40, n = 34) to 210 μs (± s.d. 110, m = 123). The reversion of Δt between 7 and 10 days to about 130 μs (± s.d. 50, n = 93) is intriguing.

Morphologic data. We found no changes in the mean cross sectional area of the muscle fibres as a result of 7 and 10 days denervation.

Conclusion. For the moment being we have the hypothesis that the denervation of only the m. EDL and m. tibialis anterior plays a role in our results. The positive effect of normal behaving muscles in the direct surroundings of the denervated EDL in our experimental condition differs from that, where the n. ischiadicus has been cut (2, 3, and many others).

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

IMPLEMENTATION OF THE APPROXIMATED LAPLACIAN SOURCE DERIVATION FOR TOPOGRAPHIC ANALYSIS OF THE QUADRICEPS MUSCLE GROUP WHILE STANDING ON THE RIGHT LEG AND STEPPING-UP

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INTRODUCTION

Close proximity to a source is characterized by a sharp inflection in the contour of the potential field. The closer the proximity to the source the greater the change in the potential contour rate of change which is directly reflected in the second derivative of the two-dimensional surface. The Laplacian operator is the second derivative gradient of change in electrical potential at a site within an electrical field[1,2]. Consequently, the Laplacian is directly proportional to the source or sink intensity at a measurement site. If electrical potential is represented by **Q**, the Laplacian can be approximated at any point within an electrical potential field by an orthogonal two-dimensional second order spatial derivative of the potential gradient symmetrical about that point in the field and represented by the formula $\Delta^2 Q = \delta^2 Q/\delta x^2 + \delta^2 Q/\delta y^2$. If a symmetrical array of electrodes records electrical potentials over a two-dimensional surface, the average potential gradient per unit distance directed to each electrode from the surrounding electrode sites is proportional to the Laplacian. Mathematically, if **0, 1, 2, 3** and **4** represents positions over a two-dimensional surface, with sites **1** and **3** positioned symmetrically about point **0** along one coordinate and with sites **2** and **4** positioned symmetrically about point **0** along a coordinate orthogonal to the first with distances of **1, 2, 3,** and **4** from **0** represented respectively by **d01, d02, d03** and **d04** and electrical potentials at each site represented by **Q0, Q1, Q2, Q3** and **Q4**, then the Laplacian, $\Delta^2 Q \sim \{[(Q0-Q1/d01)+(Q0-Q3/d03))]+\{[(Q0-Q2/d02)+(Q0-Q4/d04)]\}$. If all distances between site 0 and the other sites are equal to 1 unit distance this reduces to **(4Q0-Q1-Q2-Q3-Q4)**. If more electrodes surrounding the target site are used and/or the distances between surround electrodes to target site vary, the estimate of the Laplacian is calculated in the same manner to derive the net average potential difference directed to the target site calculated by summing the individual voltage differences each weighted inversely proportional to distance from each position to the target site. If the sampling over the two-dimensional area is non-uniform, a scaling factor proportional to the number of samplings per unit area must be introduced to derive an estimated Laplacian that is equivalent across the two-dimensional space. In the above example, if the unit area for measurement is uniformly a 1 unit distance surrounding each target site, then the Laplacian estimate can be scaled by the total number of distance measurements taken within the area—four in the example above.

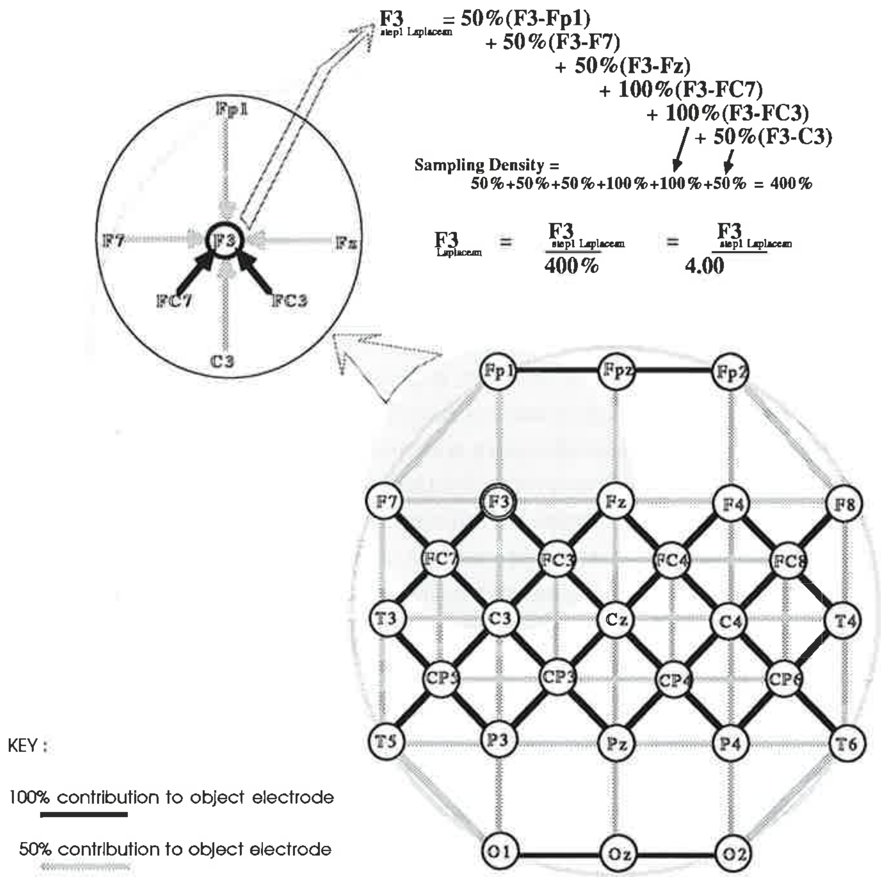

Fig. 1. Electrode reference configuration for 29 electrode montage (extended EEG International 10-20) showing the derivation of the approximated Laplacian transformation for each scalp position as the difference between the object position value and the weighted average of the surrounding positions. The electrode reference configuration was applied directly to the skin surface of the thigh over the quadriceps muscle group.

METHODS AND PROCEDURES

The implementation of Approximated Laplacian Source Derivation (ALSD) was performed by computer off-line following procedures outlined by Hjorth[1] and by Nunez[2], but modified to estimate ALSD at "edge" electrode sites (e.g.., F7, T3, T5, O1, Oz...F8, Fp2, Fpz—International EEG 10-20). Figure 1 above provides a diagram of the approximated Laplacian transform algorithm for electrode site F3 as well as positions and reference links of other sites. The units of measure for ALSD are υVolts/10cm^2 which indicates in υVolts the potential field gradient change at the target site on the recording surface with respect to all surrounding measurement

sites within a 10 cm^2 area as depicted in Figure 1 for F3. At edge electrode sites, ALSD was computed based on the surrounding half-field of sites; for T3 this would include sites F7, FC7, C3, CP5, and T5. This treatment of the edge sites is equivalent to adopting the assumption that when a bio-electric generator source is near an edge electrode, the resultant potential field beyond the edge of electrode sites is approximately a mirror image of the field adjacent to that edge. For sources located along the edge and oriented either radial or tangential to the circumference of the sphere (scalp) or cylinder (thigh), this assumption is basically correct.

Several analytical techniques which enhance electrophysiological measurement precision were applied to EMG data: (1) a dense array of 29 surface recording electrodes was applied following expanded International 10-20 EEG electrode placement system; (2) EMG data was Laplace transformed and then spectral analyzed; and (3) topographic images of Laplace transformed EMG spectral data were generated. All recordings were made in an electrically shielded room. For study 1 the quadriceps muscle group (QMG) was recorded while standing with full weight on the right leg. Study 2 evaluated concentric contraction of the QMG during stepping-up onto a 22 cm high platform with full weight on the right leg.

The subject was a 17 year old male with no history of injury to his lower extremities or knee. The recording field consisted of 29 surface electrodes placed on the right thigh. The proximal edge of the field was approximately at the head of the femur, with the distal edge located slightly above the patella. The recording field extended to the medial and lateral edges of the anterior surface of the thigh. The goal of the electrode placement was the ability to monitor from Rectus Femoris (RF), and Vastus Lateralis (VL) and Medialis Oblique (VMO). The reference electrode was placed on the medial/proximal aspect of the tibia resulting in a monopolar electrode placement montage. EMG was amplified, monitored and recorded on hard disk using a Nicolet Pathfinder I physiograph. The Laplacian transformation and spectral analysis was performed off-line using 1.25 second epochs. Spectral topographic images were generated with Nicolet software and hardware.

RESULTS

Study 1: The topographic image revealed a strong source generator in the 80-200 Hz frequency range extending over the length of the RF. Very little power was noted below 80 Hz. For the VMO no activity occurred except at the extreme distal end near the knee, while VL activity only occurred proximally. A comparison of the raw EMG spectra with the Laplace spectra demonstrated the ability of the transformation to isolate the source generator. For the raw data, an area of increased and undifferentiated EMG spectral energy spanned a region which topographically corresponded to the proximal half of the VMO and of the RF.

Study 2: The images revealed source generators along the length of the VMO and VL, both of which were strongest near the knee and along the entire length of the RF. Maximal VMO power occurred in the spectral range of 15-95 Hz, peaking in the 60-80 Hz frequency range immediately proximal to the knee, whereas VL's peak energy range was 40-80 Hz with the major RF energy spectrum at 40-200 Hz. Most of the power from VL was also concentrated just proximal to the knee. A second region of increased power appearing either over the proximal portion of the RF or possibly from the Tensor fasciae latae was also observed in the 40-95 Hz range. Of the vasti, the greatest power was recorded from the VMO immediately proximal to the knee. For stepping-up the activity of RF is identical to that for standing on one leg, but with a 2 to 3 fold increase in energy level. Unlike the spectral frequency range of the VMO and VL which appear to have a spectral edge near 80 Hz, the RF continued to be active up to 200 Hz. The observed increase in RF power while stepping up, as opposed to standing on one leg, probably reflects the relative increased load on the femur while lifting the body up to a stand position. In these images, muscle groups appear to be clearly distinguishable by location and differential energy spectra. A clear delineation of the boundaries of the VMO from the VL could also be seen immediately proximal to the knee between 15 and 80 Hz.

CONCLUSIONS

The generated images have face validity with respect to known structure and function of the QMG under the specified maneuvers. For example, the observation that both vasti showed maximal power immediately proximal to the knee during the step-up is not surprising since both serve to stabilize the knee, with findings for the VMO being consistent with the known function of the VMO toward the end of extension in prevention of lateral dislocation of the patella induced by the lateral displacement created by the activity of the VL. Moreover, the topographic location of greatest VMO power was detected by the electrode field over the lowest and most distal part of the VMO where the muscle bulges most prominently. There was virtually no cross-talk between the RF and adjacent muscle groups RF, VL and VMO. Topographic images of spectral Laplace transformed EMG from the thigh while standing on the right leg allowed spatial identification of the RF as the principle source generator of EMG. The observation of little or no power below 80 Hz while standing and for stepping-up appears to be unique finding for RF. Topographic images of raw EMG provided a very different image from that obtained with the Laplace transform, confirming that conventional surface EMG recording using monopolar referencing can be misleading. As a final note of caution, because the results are reported from a single subject, issues of individual variability have not been addressed, hence some of the novel findings may be idiosyncratic to the subject.

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TOPOGRAPHIC MAPPING OF SURFACE EMG OF QUADRICEPS THROUGH A MODIFIED LAPLACEAN SOURCE DERIVATION

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INTRODUCTION

A major constraint for understanding muscle electrophysiology and function is the precision with which muscle activity can be measured. Traditional procedures for the assessment of electromyographic activity have entailed the insertion of fine wire electrodes into the muscle or differential amplification of attached surface electrodes[1] . The placement of the needle electrodes records from a very small and specific set of EMG source generators, and may not provide an adequate representation of "total" muscle function. As Wolf et. al.[2] have recently shown, separate portions (or compartments) of the muscle can be differentially activated during the same movement. Volume conduction from adjoining muscles or motor units provides a major source of noise when attempting to isolate energy from a particular EMG source generator. Recently, spectral analysis of EMG activity has demonstrated its value for quatification of EMG information[3] . What this study proposes to do is to quantify electromyographic activity using a topographic mapping technique used in EEG analysis[4,5].

METHODS AND PROCEDURES

Several analytical techniques which enhance electrophysiological measurement precision were applied to EMG data: 1) a dense array of 29 surface electrodes was applied following an expanded international 10-20 EEG electrode placement system; 2) EMG data was Laplace transformed to eliminate signal spatial dispersion due to volume conduction[5]; 3) EMG transformed data was spectral analyzed; and 4) topographic images of Laplace trans-formed EMG data were produced to facilitate evaluation of the EMG activity over space, function and energy spectrum.

Figure 1 displays the implementation of the Laplace transform. This transform is particularly well suited to delineate bioelectric activity at the edge of recording boundaries. In essence, it allows the electrode array to operate in a fashion similar to that of the retina. For example, only those electrode sites which are electrically different from their surrounding electrodes will "light up" on the topographic display.

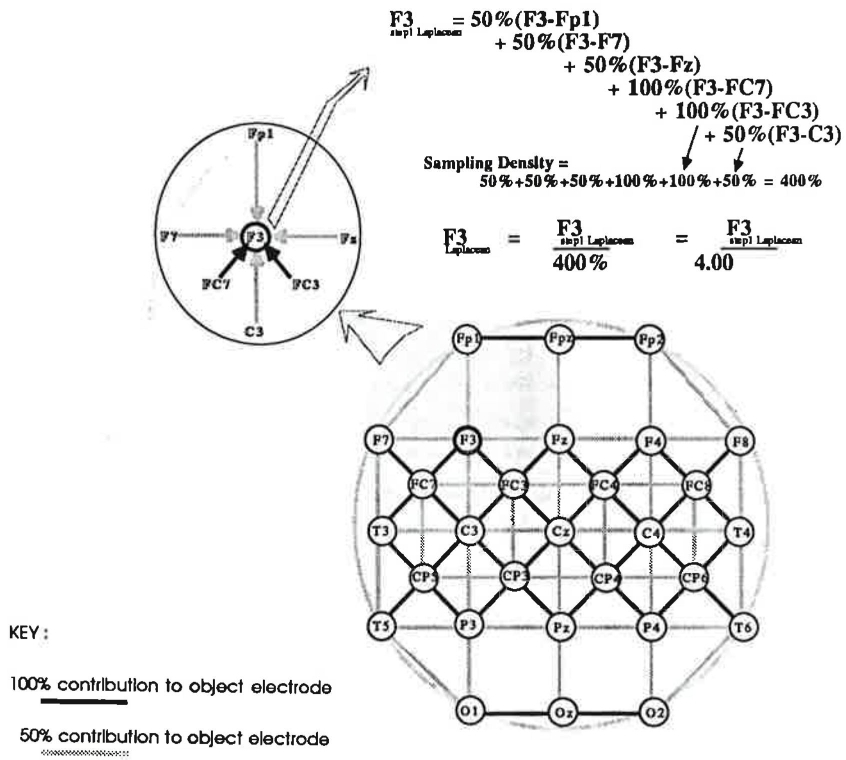

FIG 1. Electrode configuration of the 29 electrode montage showing the derivation of the approximated Laplacean transformation.

The subject was a 17 year old male with no history of injury to his lower extremity. The recording field consisted of 29 surface electrodes held in place on the right thigh with collodion. The proximal edge of the field was at the head of the femur, with the distal edge located slightly above the patella. The recording field extended to the medial and lateral edges of the anterior surface of the thigh. The goal of the placement was to monitor from Rectus Femoris (RF), and Vastus Lateralis (VL) and Medial Oblique (VMO). The reference electrode was placed on the medial/proximal aspect of the tibia resulting in a monopolar electrode placement montage. The EMG was amplified, monitored and recorded on hard disk using a Nicolet Pathfinder physiograph. The Laplacean transformation and spectral analysis was performed off-line using 1.25 second epochs. Spectral topographic images were generated with Nicolet software and hardware.

Two movement patterns were studied. The first was the closed kinetic chain of eccentric and concentric contractions during squatting while standing on the right leg. The second was a study of maximum voluntary contraction (MVC) and 50% MVC of the quadriceps at 0, 30, 45, 60 and 90 degrees of extension. The subject was in the supine posture, but the thigh was not strapped down.

RESULTS

The topographic images generated during the concentric phase of squatting indicate source generators near the knee in the most distal regions of VMO and of VL, along with a proximal source in the area of RF. The VMO is most clearly imaged in the spectral range of 40-95 Hz, VL in the 40-60 Hz range and RF in the 40-80 Hz range. See Figure 2.

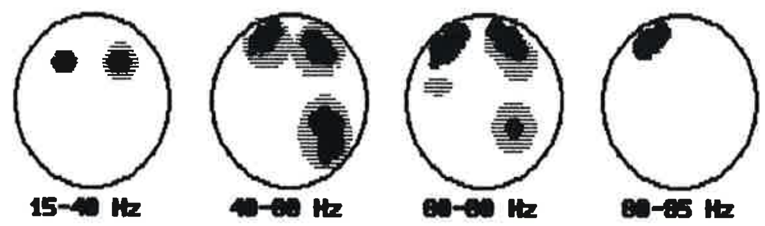

FIG. 2. Topographic display from quadriceps muscle during concentric phase of squatting. Top = knee, left side = medial aspect. Numbers represent frequency range of the spectrum plot. The darker the area, the greater the level of EMG activity.

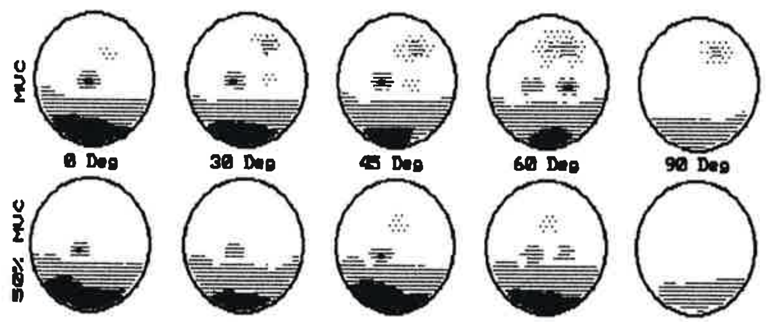

Fig. 3. Topographic maps of quadriceps muscle during MVC (top) and 50% (bottom) isometric contraction. The darker the area, the greater the amount of EMG activity.

The findings for the isometric contractions indicate that during MVC, the energy seen in the topographic display shifts from the proximal end to the distal end as one goes from 0 degrees of extension to 90 degrees. The maximum distal energy was noted in the area of VL and VMO at 60 degrees. The findings for the 50% MVC show maximal energy being displayed in the proximal portion only. See Figure 3.

DISCUSSION

The topographic maps of approximated Laplacean transformed EMG show a startling ability to image myoelectric potentials in two dimensional space. The findings for the concentric phase of squatting show a clear delineation of the individual muscles, especially VMO and VL. One can clearly see the separation of these two opposing muscle groups, as they stabilize the knee during this coordinated movement[6]. Cross talk is not a problem. The RF may also be imaged, but only at its proximal section and only at the slower frequencies.

With isometric contractions, the topographic imaging shows a large involvement in what would appear to be hip flexors, such as Iliopsoas. This finding is understandable, given that the thigh was not strapped down during the procedure. The finding of the distal portion being more active during 45 and 60 degrees of flexion should be of theoretical importance.

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Electromyographical Kinesiology
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PATTERNS OF ELECTROMYOGRAPHIC ACTIVITY IN THE HUMAN LATERAL GASTROCNEMIUS MUSCLE DURING WEIGHTBEARING AND NON-WEIGHTBEARING TASKS

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INTRODUCTION

Histochemical and EMG analyses on cat lateral gastrocnemius muscle (LG), have demonstrated the presence of specific subvolumes or divisions of muscle which are innervated by a single primary nerve branch of the peripheral nerve to that muscle. These subvolumes have been called neuromuscular compartments and are thought to contain a unique population of motor units.[1] For studies of human two-joint muscles, however, we have only begun to define the neural distribution while not yet clearly comprehending the patterns of muscle activity.[2,3]

As a result of extensive microdissections from cadaver material, we have reason to believe that at least four distinctive partitions exist within the human LG. The purpose of this study was to determine if the compartmentalization concept described for cat LG muscle may also be applied in human LG muscle at these four predetermined sites during functional weightbearing and non-weightbearing tasks.

MATERIALS AND METHODS

Volunteers (n=20) from the Emory University Community, mean age 25.9 years, range 22-40, randomly performed three trials of the following tasks: plantar-flexion in prone, knee flexion in prone, knee flexion with plantarflexion in prone, plantarflexion standing, knee flexion with plantarflexion in standing, progressing up a step with the test leg stationary on top of the step, and progressing up a step with the test leg stationary on the floor.

In a one, two hour experiment, each subject was first prescreened, then had bipolar, fine-wire electrodes (25 u m uninsulated tips), inserted into four areas of the LG using the technique developed by Basmajian.[2] The exact locations of insertion were based upon a morphological proportional method derived from previous measurements on cadaver legs.[3]

EMG was preamplified at the skin surface (x1000) and led to: an amplifier, four channel oscilloscope, FM tape recorder, and computer for simultaneous analog to digital conversion of the amplified and rectified EMG signal. Subjects first performed a maximum isometric LG contraction with the ankle plantarflexed and the knee extended as a bases for calculating normalized EMG during specific movement tasks. Spiked-triggered averaging of isolated single motor units using two additional subjects to determine the presence of time locked activity between locations across tasks. Spiked- triggered averaging is a process in which an EMG event at one location is correlated with any recorded events occurring at other locations, before, during or after the original event.

RESULTS

Means and standard deviations were computed for EMG activity for each task. For the task of unilateral plantarflexion in standing, greater amounts of activity were observed from all of the four electrodes when compared to all other tasks (mean normalized EMG activity 9.87, and 8.36). When progressing up a step with the non-test leg, with the test leg stationary on top of the step, a greater amount of activity was seen from the distal lateral electrode (mean normalized EMG activity 55.34), than from each of the proximal electrodes (mean normalized EMG activity, medial electrod 35.17, lateral electrode 31.92). During knee flexion in standing and knee flexion in prone, similar patterns of EMG activity were observed. Each distal electrode (mean normalized, medial electrode 20.38, lateral electrode 17.54 and medial electrode 17.68, lateral electrode 14.66), and the proximal medial electrode (mean normalized EMG activity 19.1, 15.38), recorded a greater amount of activity than the proximal lateral electrode (mean normalized activity 9.87, 8.36). No consistant pattern of activity was found from all locations across all tasks.

Two-way four factorial ANOVA's for repeated measures were performed for normalized data. Significant differences ($P < .05$) were found: within all locations between all tasks, and within all tasks between all locations except for : knee flexion with plantarflexion in prone, unilateral plantarflexion in standing, and progressing up a step with the test leg on the floor.

Spiked triggered averages were assessed for 64 isolated motor units. No evidence of time-locked activity across two or more sites was observed in 79% of our observations; thus suggesting that the analyses from these sites represented differential activity.

DISCUSSION

A primary finding indicated more activity was observed from all electrodes during plantarflexion in standing than during all other tasks. This observation emphasizes the importance of the LG as a primary plantarflexor of the ankle and as a stabilizer of the ankle joint during weightbearing.

The task of progressing up a step, with the test leg stationary on top of the step, was the only task requiring active lengthening of the LG. During performance of this task, the distal electrodes recorded a greater amount of activity than each proximal location. This finding of differential activity noted when LG is actively lengthening is consistant with findings in studies on the cat sartorius muscle under lengthening conditions of the gait cycle.[4] This differential activity was explained by proposing a concept of task specific groups of motor units which, for our purposes, can be considered the equivalent of neuromuscular compartments. A specific group of motor units producing a greater amount of activity in the distal electrodes during lengthening tasks, may explain the differential activity levels observed during this task. The fact that no differential activity occurred between locations when the LG was actively shortening at two joints simultaneously, suggests that muscle activity may be required at all locations under these conditions.

During knee flexion in prone and knee flexion in standing, the proximal lateral electrode consistantly recorded a smaller amount of activity when compared to the three remaining locations. This differential activity noted in knee flexion tasks may be due to the muscle fiber orientation with the LG. Gans has suggested that fiber orientation has an effect on force production.[5] The muscle fibers in the human LG may be oriented in such a manner that the distal fibers and the proximal medial fibers are in a more biomechanically advantagous position than the proximal lateral fibers during knee flexion. For these tasks, the same pattern of muscle activity at each of the four locations within the muscle was observed. This finding suggests that body position and weightbearing status of the opposite extremity may not affect LG muscle activity during knee flexion.

When reviewing the activity levels recorded at all locations across eight tasks, there was no specific pattern of EMG activity from all electrodes, regardless of the nature of the tasks. This finding implies that EMG activity is not uniform across tasks for individual LG locations, but rather may be unique to individual tasks. This observation supports the idea of task specific motor unit groups, or neuromuscular compartments, existing in the human LG muscle.

Data from this study emphasize the importance of the LG as a primary plantar-flexor of the ankle. Task specific patterns of muscle activity may occur within the human LG muscle: during knee flexion activities, independent of body position, independent of weightbearing status of the opposite extremity, and when the LG is actively lengthening. Differential activity between locations may not occur when the LG is acting at two joints simultaneously.

If future studies regarding muscle partioning do validate the existence of neuromuscular compartments, then these results have clear clinical implications. Patient rehabilitation time may decrease and the quality of treatment may improve through a more specific application of modalities, such as: biofeedback and electrical stimulation. More importantly, this knowledge base would allow therapeutic exercises to be either more task specific or more clearly and justifiably deliniated.

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KINESIOLOGY

Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

MUSCLE COORDINATION AND PERFORMANCES IN ANTERIOR CRUCIATE LIGAMENT (ACL) -INJURED SUBJECTS

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INTRODUCTION

Quantitative analysis of the electromyogram (EMG) has been used as a method to understand the way in which muscle activity and coordination can be changed to increase the functional stability around the knee in ACL-subjects. We have reported on altered muscle-coordination in ACL-subjects, compared to controls (1,2). The results, reported here, show an earlier onset of EMG-bursts for muscles around the knee in patients, especially in the lateral hamstring and medial gastrocnemius muscles. The duration of the EMG-burst also tended to be prolonged in patients.

Comparing patients with poor and normal knee functions, we report on significantly earlier onset-times of EMG and longer durations of bursts in the medial gastrocnemius muscle in patients with normal knee function.

We conclude that changes in the coordination of the muscles around the knee is useful to restore normal knee-function in ACL-patients.

MATERIALS AND METHODS

We compared 14 patients (20-35 years old) with isolated, total ACL-ruptures, arthroscopically verified 2-3 years previously, with volunteers that matched the patients as regards sex and age. Some of the patients suffered from giving-way symptoms in the affected knee during daily activities or recreational sport. The volunteers had no history of knee-injuries.

We used a performance-test, adapted from Tegner et al. (3), to evaluate the degree of dysfunction in the ACL-patients. The test included a one-leg hop, running in a figure of eight, running up and down a staircase. The maximal score was 100. We compared seven well-functioning ACL-patients, with a mean score of 95 (range 84-99) with seven ACL-patients with a poor function and a mean score of 60 (56-68).

To investigate if there is any neuromuscular correlation to the different performances in the two groups of ACL-patients, we investigated the EMG from the muscles around the knee during walking on a treadmill. We recorded EMGs from m. Vastus Lateralis (VL), m. Vastus Medialis (VM), Lateral Hamstring (LH), Medial Hamstring (MH), m. Gastrocnemius Medialis (GM), m. Tibialis Anterior (TA), using bipolar surface electrodes (with a 2.0 cm spacing), which we placed along the muscle. EMG-signals were amplified (Dias 15C01) and filtered (20-500 Hz) by a first-order bandpass filter. We measured the contact between the heel of the shoe and the ground by a pressure-detector placed under the heel of the patient's shoe.

We digitized EMGs and ground contact at 512 Hz on a personal computer, which displayed raw data in real time.

All EMG-signals were analyzed and divided into step cycles of 100% with respect to heel contact (4). We found the onset-time of the ground contact, using a method of successive integration (5). The EMG onset- and offset-times were automatically detected, noting the variance of the EMG-signal and we found the duration of an EMG-burst by subtracting offset- and onset-times (6). By rectifying and filtering the EMG-signal, we calculated the EMG-profiles for each muscle (4). The amplitude of the profile was normalised with respect to the mean value, and a mean EMG-profile was described by a mean and standard deviation curve from about 10 step-cycles. We calculated the power of the EMG-signals was as the Root Mean Square (RMS). We then elevated the treadmill from a horizontal level to 11.25 degrees (with a 25% incline), and asked the patient to walk for 3 min. at 5 km/h. The Onset-time and duration of each burst within the walking cycle and the EMG-profiles were calculated at 0, 5, 10, 15, 20 with a 25 % incline.

RESULTS

Muscle-coordination in normal patients and ACL-patients

When we increased the inclines, the amplitude of the EMG-profiles also increased for all recorded muscles in normal and ACL-patients. In both groups, we found a significant ($p<0.05$) shift in the time for the peak activity at inclines of 10% and higher in the HL, and of 15% and higher in the HM. The time to the peak of the EMG-activity in the HL and HM appeared before heel contact at horizontal walking and after heel contact at 25% uphill-walking. We found a shift in the peak time in the GM at 25% uphill-walking in controls and a significant shift at 15% and higher in ACL-patients. No significant changes in the peak time were found in the VL and VM.

The onset-times of the EMG were significantly earlier in ACL-patients in the VM, VL, HM, HL and GM-muscles at all inclines, when compared to the controls; the most pronounced differences were found in the HL and GM.

In ACL-patients, we found that the durations of the EMG-bursts were significantly longer in the VL, VM, HL, and GM-muscles at all inclines, when compared to the controls. The duration of the burst in the GM-muscle was 35% (+/-6%) of the step-cycle in controls and 38% (+/-7%) in ACL-patients.

Muscle-coordination in ACL-patients with a good and poor functional stability

Table 1 shows the onset times and duration at a 0% and 25% incline for ACL-patients with high and low score, and a normal control group.

Incline	0%		25%	
	Onset	Duration	Onset	Duration
High score	5+/-5	43+/-4	3+/-2	51+/-3
Low score	15+/-7	34+/-8	12+/-5	41+/-4
Control	15+/-4	35+/-6	10+/-4	42+/-4

Table I: EMG onset-time and durations of bursts in percent of a step-cycle for the GM-muscle. +/- 1 SD is shown. "High score" equals a mean score of 95, and "low score" a mean score of 60 (see Methods).

A comparison between the onset-times and the durations of the burst shows a significantly earlier onset-time and longer duration for the GM-muscle in ACL-patients with high score than in ACL-patients with a low score or when compared to a normal control group. We found no differences in the onset-times and durations for the other muscles.

Further, we found that the RMS was also significantly increased at a 25% incline in the high score group, when compared to the low score group.

DISCUSSION

ACL-patients with a low score have a muscle-coordination similar to controls, whereas ACL-patients reaching a high score have altered muscle-coordination. The findings indicate that an improved functional stability might be a result of the altered muscle-coordination around the knee, when measured from the performance-test by Tegner el al. (3). The surface-EMG, measured from the GM muscle, indicates that this muscle plays an important role in the improvement of the functional stability after an ACL-rupture.

We believe that the increased onset-times and durations of bursts of the EMG in ACL-patients with a high score compensate for the lost resistance, normally obtained by the anterior cruciate ligament during extension of the knee just before heel contact. The changed muscle-coordination increased the stiffness around the knee, which protect against unexpected perturbations during the extension of the knee.

Using the present technique, it is possible to study where and how, during a step cycle, the muscle coordination differ between normal and ACL-injuried subjects. This technique is now used to develop new rehabilitation programmes for ACL-injured subjects.

ACKNOWLEDGEMENTS

This work was supported by Team Danmark and Assurandør Societetet, Denmark.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

ELECTROMYOGRAPHIC AND KINEMATIC PROFILES DURING DIFFERENT GAIT VELOCITIES AND INCLINES

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INTRODUCTION

The strategies by which the CNS control various gait conditions can be studied non-invasively by the electromyographic (EMG) and the kinematic profiles. The EMG profiles reflect the sum of neuronal output (firing, recruitment) to the muscles and the kinematic profiles provide information of how this output is converted into movements. The neurological mechanisms which control the gait velocity are unclear, and the purpose of the present study was to investigate how the central nervous system controls the gait velocity of walking on horizontal and sloping grounds.

MATERIAL AND METHODS

Ten healthy young subjects participated. All volunteers walked on a treadmill for 30 sec under the following 4 conditions: 3.5 km/h, 0% elevation; 5.0 km/h, 25% elevation; 3.5 km/h, 0% elevation; 5.0 km/h, 25% elevation. The order of the tests was randomized. A pressure-sensitive transducer was mounted externally under the heel to measure the heel contact. The knee flexion/extension angle was measured by a triaxial goniometer and the flexion/extension angular velocity and acceleration profiles were computed. The EMGs were recorded from the right m. gluteus maximus, m. biceps femoris, m. vastus lateralis, m. gastrocnemius lateralis, and m. tibialis anterior. EMGs were pre-amplified ten times by small headstages (model 2010, Intronix Technologies, Ontario, Canada), transmitted to the amplifiers, and filtered (20-500 Hz) before data collection. Prior to the profile calculation the EMG was full wave rectified and smoothed. The mean profile amplitude and the stride time were normalized to 100% (1). The EMG and kinematic profiles were averaged on the basis of 20 step cycles. Wilcoxon's signed rank test and Mann-Whitney U-test were used for statistical analysis, and $P<0.01$ was regarded as significant.

RESULTS

Velocity changes: Changes in gait velocity did not change the overall patterns (table 1) of the kinematic and EMG profiles (fig. 1, 2), whereas the peak am-plitudes of the profiles (table 2) changed. The knee angular acceleration profile shows a strong deceleration of the flexion in the period 92-95% of stride, with a peak amplitude related to gait velocity. This deceleration cor-responds to the peak EMG in biceps femoris which occur around 87% (fig. 2). The delay between the peak EMG and the peak deceleration corresponds to approximately 100 ms, corresponding to the delay between EMG and force.

TABLE I.
PROFILE PATTERN CHANGES.
The changes in the profile patterns for increased gait velocity and incline. For the 'no change'-condition the amplitude values may change but the overall pattern remains constant.

	3.5km/h to 5km/h	0% to 25%
Flexion/extension(^{o})	No change	Changed
Angular velocity (^{o}/s)	No change	No change
Angular acceleration ($^o/s^2$)	No change	No change
Gluteus maximus	No change	No change
Biceps Femoris	No change	Changed
Vastus Lateralis	No change	No change
Gastrocnemius Lateralis	No change	No change
Tibialis Anterior	No change	Changed for 3.5 km/h

Incline changes: When the incline increased, the flexion/extension angle profile was changed around heel strike, as the knee joint was less extended than in horizontal walking. In the period around push-off, the knee angle was around 20 degrees under all conditions, which is very adequate as the hamstring muscles develop maximum knee moment for a 20 degrees knee angle (2). The changes in the flexion/extension profile did not affect the pattern of the angular velocity and acceleration profiles (table 1) but only the velocity and acceleration peak amplitudes (table 2). It is remarkable that the angular velocity and acceleration profile patterns remained constant as the peak EMG amplitude in some cases changed substantially.

TABLE II
PROFILE PEAK AMPLITUDE CHANGES
The mean increases (+) in the peak amplitudes of the profiles as the gait velocity and incline are increased.

	3.5km/h to 5km/h	0% to 25%
Flexion/extension (^{o})	+5^o for 0%	+40^o
Angular velocity (^{o}/s)	100^o/s	150^o/s
Angular acceleration ($^o/s^2$)	2000$^o/s^2$	2500$^o/s^2$
Gluteus maximus	+150%	+2000% for 3.5km/h +1000% for 5.0 km/h
Biceps femoris	+200%	+500% for 3.5km/h +750% for 5.0km/h
Vastus lateralis	+150%	+1000%
Gastrocnemius lateralis	+200%	+1000%
Tibialis anterior	+100% for 0% +450 for 25%	+300% for 3.5km/h +900% for 5 km/h

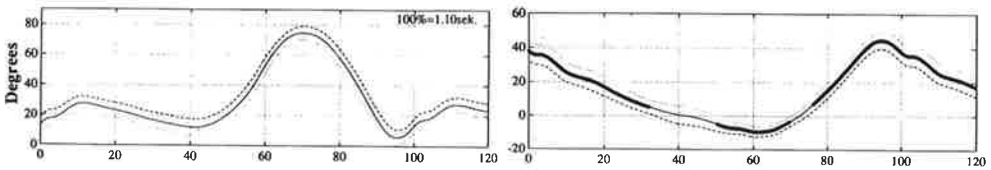

Fig. 1. The knee flexion/extension angle profile (left) for 5km/h uphill-walking. The mean (full line) and standard deviation (dotted lines) are calculated for 10 subjects. The difference profile in degrees between 5km/h uphill- and 5km/h horizontal-walking (left). The bold line on the difference profile indicates significant (P<0.01) difference. Heel strike is 100%.

DISCUSSION

An increased gait velocity caused no modulation of the shape of the EMG profile patterns neither for horizontal nor graded walking. Only the amplitudes were modulated, as previously suggested (3), to occur when gait velocity was increased for horizontal walking. The neuronal mechanisms for this strategy remain unclear. Winter (4) suggested that gait velocity control was obtained by varying both cadence and force. The peak EMG activities, which may occur during increased gait velocity, can be substantially larger than what can be obtained during static maximum voluntary contractions (5). This indicates that the stretch reflex contributes to the control of stereotyped motor tasks in addition to the supraspinal and spinal control strategies for locomotion. In the present study, we did not relate the EMG activity to the activity level, obtained during a 100 % maximum voluntary contraction, as it is difficult in most cases to activate a single muscle 100% in a muscle group, and hence obtain a selective and reliable estimate of the maximum EMG activity.

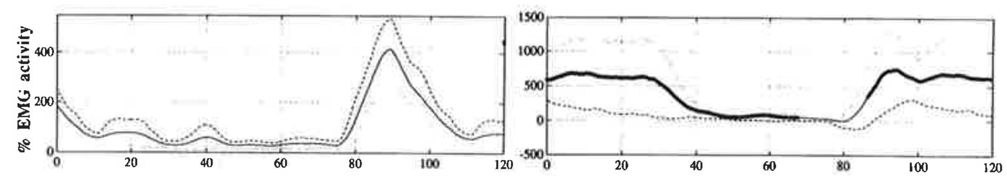

Fig. 2. The EMG profiles for m. biceps femoris for horizontal walking at 5 km/h (left). The mean (full line) and standard deviation (dotted lines) are calculated for 10 subjects. The relative differences profile between 5km/h uphill- and horizontal-walking (left). The bold line on the difference profile indicates significant (P<0.01) differences. Heel strike corresponds to 100%.

Few studies have investigated the EMG activity during walking on various inclines (6). During the late swing phase, the ankle dorsiflexors have to be activated more for uphill walking to prevent toe stubbing. The greater flexion at the knee joint in the early stance phase and the late swing phase for walking on incline, requires an increased tension in the hamstring and quadriceps muscles to stabilize the knee in this flexed position. During the stance phase, the extensor muscles of the hip (gluteus maximus, biceps femoris) and knee (vastus lateralis) assist the triceps surae muscles to provide push-off. An increase of the incline require additional power in these muscle groups. When the velocity was increased, large 'gain' differences were obtained between muscles. This may be a result of the many factors which modulate the EMG amplitude during dynamic conditions - contraction velocity, muscle length, eccentric/concentric contraction, reflex contribution, tension, EMG/torque relation. In general, an increased incline requires substantially more 'gain' than an increase in the gait speed. Interestingly, the shape of the angular velocity and acceleration profiles remained constant when incline and velocity increased, although the shape of the knee flexion/extension angle profile was altered. This suggests that either the velocity or the acceleration profiles might be a template for how the central nervous system should control and plan the strategy for various demands to change locomotion. The variability and flexibility of the muscle activity during gait is found to be substantially greater than what is seen in the effective knee torques that they produce (1), which may indicate that the CNS control the net pattern of the torque more tightly than the individual muscles. This should be further investigated as it has implications for artificial limb control.

ACKNOWLEDGEMENTS
Det Obelske Familiefond, Aalborg Stifts Julelotteri and Assurandør Societetet were kindly acknowledged for their financial support.

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Electromyographical Kinesiology
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EXCESSIVE MUSCLE FATIGUE RESPONSE; THE CULPRIT IN FIBROMYALGIA ?

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INTRODUCTION

A number of chronic back pain patients seen in our laboratory reported an accentuation of lower back pain associated with increased stiffness brought on by cold, damp weather, and a rapid onset of back pain when they were exposed to cool air drafts (air conditioning, open windows) on their backs. The symptoms seemed to be ameliorated by local heating or a hot shower, stretching exercises or moderate activity. These weather and temperature related symptoms, which are part of the overall clinical picture of primary fibromyalgia (1), have as yet not received particular attention, although they exercise genuine and consistent effects.

In light of the muscular pathophysiologies proposed as a cause of primary fibromyalgia (1,2), a study was initiated which investigated the effect of surface cooling on muscle function, particularly examining evidence of metabolic muscle fatigue, as derived by parameters of the electromyographic power spectrum.

METHOD

Five normal volunteers participated in this investigation (4 males, 1 female; age=22.4; body weight=76.7 kg, sd.=10.15; height=182.0 cm, sd.=6.4). Prior to the cooling procedure, electromyographic (EMG) power spectrum baseline data and skin surface temperatures were recorded. The details of the EMG recording procedure has been described elsewhere (3). Paraspinal constant force contractions were generated during a weight lifting trial in a reference frame in which the position of the feet, pelvis and spine could be controlled (4).

The cooling of the paraspinal muscle region adhered to the therapeutical application of cold as outlined by Wadsworth and Chanmugam (5). Crushed ice, wrapped in wet towels was placed on the back (L1-L5 region) and changed at 1-min intervals over a period of 20 min. Surface temperature dropped logarithmically from 31.7^{0}C to 8.6^{0}C. Hartvickson (6) and Bierman (7) have shown that this procedure will lower effectively the intramuscular temperature to 30-33^{0}C, depending on the distance and amount of fatty tissue between surface and muscle tissue.

Subsequent to the application of cold, the contraction trials were repeated 3 times separated by a 5 min rest period; this time period had previously been found sufficient to allow for complete metabolic recovery in paraspinal muscles (8).

RESULTS

The collected MF data were subjected to a linear regression analysis (1) for the calculation of intercepts (estimates of initial median frequencies: MF), and of regression coefficients (degree of change of the power spectrum during contraction time, indicating muscle fatigue). To reduce the complexity of the electromyographic measure and to make statements about levels (e.g. multifidus versus iliocostalis lumborum) the bilaterally recorded data were averaged.

Analysis of the estimated initial median frequencies for the multifidus and iliocostalis muscle showed a close relationship with surface temperature changes (r=.988), thus confirming earlier reports (9) of a linear and rigid relationship between initial median frequency values and intramuscular temperature. Calculating the recovery shift of initial median frequencies for the cold tests (expressed as the percentage change between baseline and first post-cooling test run), revealed that both muscles recover comparably over time, although the recovery was still incomplete at the 3rd post cooling test (about 41%), which was recorded 14.0 min after the termination of applied cold to the lower back area.

The accompanying muscle fatigue data for each cold trial, expressed as percentage increase relative to the recorded (pre-cooling) baseline data, are shown below. Overall, one can notice a differential increase of the fatigue rate in the multifidus and iliocostalis over post-cooling assessment trials. For the multifidus muscle, the fatigue rate increases steadily and exceeds the initial baseline value by 63% at the last contraction trial. The increase is more pronounced in the iliocostalis. This muscle, located close to the surface, shows an increase in the fatigue rate which exceeds the baseline value by 167% at the 2nd post-cooling run with the tendency to drop off at the succeeding contraction trial.

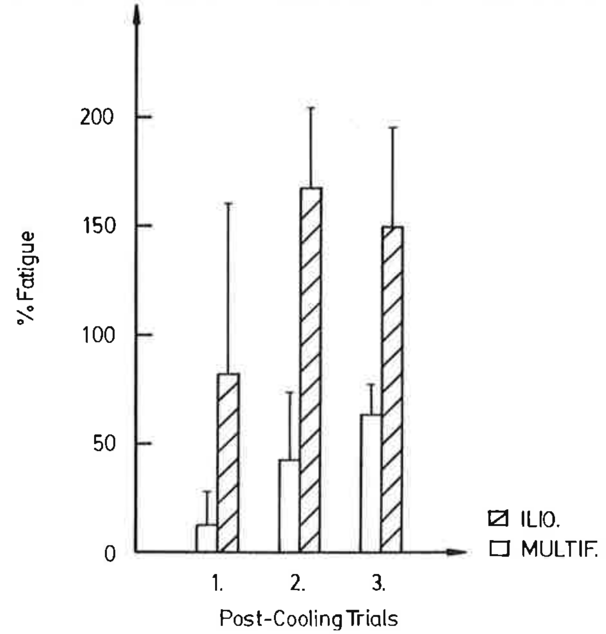
DISCUSSION

In addition to confirming earlier observations of a close relationship between initial MF values and temperature (9), our data revealed an accumulative tendency in the muscle fatigue parameter over the three post-cooling paraspinal constant force contractions. Although the initial MF estimates indicate gradual recovery of the muscles from cooling, the highest

muscle fatigue rates were not observed while the apparent intramuscular temperature was at its lowest (as indicated by the MF values), but at succeeding contraction trials. These increase in fatigue rates over contraction trials may be due to either an accumulation of acid by-products because of insufficient supply of O_2 (10), or insufficient removal of metabolic by products because of reduced muscular blood flow (11,12); both mechanisms would lead to increased acidity in the muscle tissue which eventually would impede muscle contractions and would make it increasingly difficult and painful to activate the muscle in question (13).

Although research on the effect of overall lower body temperature has been found to improve endurance performance in test subjects (14,15,16), its mechanism has been related to a delay of discomfort from increasing body heat which counteracts the subjects' drive to sustain physical activity, but not to improved functioning of the muscles involved. In fact, precooling resulted in reduced muscular performance (i.e. pedal rate; 16) and was related to cardiovascular and thermoregulatory control mechanisms.

The effect of surface cooling in Fibrositis patients has, as yet, not received particular attention with regard to electromyographic responses. Given our findings, together with the persistent claim of sensitivity to cold with associated muscular pain in Fibrositis patients, we propose that such patients may experience excessive vasoconstriction in some muscles as an adaptive cardiovascular response to changes in surface temperature. We are presently exploring the means to investigate this hypothesis further.



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THE DEFINITION, DESIGN, IMPLEMENTATION AND USE OF EMG SOFTWARE FOR THE ACORN ARCHIMEDES 440 MICRO-COMPUTER

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INTRODUCTION

The objective of this presentation is to report the development and use of EMG software which forms part of a comprehensive sports biomechanics software package written by the author for use with his laboratory's Archimedes 440 microcomputers. These incorporate a 32 bit RISC architecture microprocessor and use pipelined execution of the three cycle instructions (fetch, decode, execute). With a clock rate of 8MHz and a very short interrupt latency of considerably less than 1 ms, this makes for a very fast machine.

The development of the EMG software was undertaken because of the lack of commercially available EMG software for this microcomputer. It was intended to be used in the laboratory's extensive research programme and in the teaching of undergraduate courses in sports biomechanics. The software obtains 8 channel EMG signals from MIE 1000 or 8000 gain skin mounted pre-amplifiers, with or without telemetry (MIE MT8) and further amplification. Signals are analogue/digitally converted (Microlink 12 bit A/D converter) and transferred to the computer using the IEEE bus at sampling rates per channel of up to 3000 Hz.

The overall software package was developed by adapting the principles of good software engineering as described by Sommerville (9) to a single person project. The most important software attribute was considered to be maintainability as there will undoubtedly be adaptations to changing requirements during the operational life of the package, and these adaptations may be the responsibility of someone other than the author of the existing package.

REQUIREMENTS DEFINITION

Because of the non-criticality of the project and its development by one person, a formal specification was not considered necessary. However, the requirements of the package were clearly defined using natural language and in a format which ensured the traceability of the requirements through the later stages of design and coding. The requirements definition included both the functional requirements of the system, i.e. the services expected of it by its users, and non functional requirements which defined any operational constraints. The functional requirements were based on a comprehensive search of the relevant literature and consultations with potential users.

Informally stated, the EMG software provides EMG data acquisition and storage and analysis, as recommended by e.g. (2 and 5), incorporating mean rectified emg, integrated emg and differentiated emg, as well as frequency spectrum analysis.

SOFTWARE DESIGN

Because of the nature of the overall package, an essentially top down functional design strategy was used with different levels of abstraction. This was undertaken to ensure that the design captured the requirements and to formally document each component of the software so that it could be easily implemented. The design proceeded through the use of data-flow diagrams as specified by Constantine and Yourdon (4). These were supplemented by a data dictionary to represent all data and process definitions. Control flow was modelled by finite state machines. The partitioning of the data flow diagrams was stopped when the processes represented were easily described by a single page minispec with a clearly defined function or related group of functions.

The user interface was designed to suit the needs and abilities of the users, to be consistent throughout the package and to be visually attractive. The Archimedes 440 window manager provides a versatile interface. However, novice users, such as many of the laboratory's undergraduate students, lack familiarity with windows environments and would need to look up instructions on dragging, selecting, scrolling etc. before being able to use the system.

For this reason, a simple menu system was preferred to the windows environment available on the Archimedes. This was designed to allow for two simple types of user response:

* a choice from a list of options using the mouse to point to and select the required option;
* very short text inputs from the keyboard e.g. to define a file name.

In both cases, it was made impossible for the user to make inappropriate choices.

The design was thoroughly tested by the use of manual checking in an attempt to ensure that all requirements had been dealt with and that the design was consistent and feasible. This also ensured that the coupling between program modules was loose and that the cohesion within each program was functional i.e. that each part of the program performs a clearly defined function. These attributes are considered to be important in good software engineering, particularly to aid maintainability.

IMPLEMENTATION AND VERIFICATION

The software was coded from the design in BASIC V for maintainability reasons, in so far as BASIC is the language most frequently used by, and familiar to, sports biomechanicians in the UK. Whilst not providing much of the data typing of e.g. Pascal, BASIC V does provide all of the control constructs and many of the parameter passing facilities of the latter language. There are no great speed disadvantages in the use of BASIC V as opposed to Pascal or FORTRAN on the Archimedes 440.

The coding was undertaken by writing structured programs, i.e. ones that conform to the definition proposed by Linger et al. (6), the control flow in which can be represented by a proper flowgraph. This approach was adopted to facilitate verification and validation of the software. Implementation dependent features of the programs were confined, wheever possible, to a few modules.

The programs were firstly reviewed by detailed desk checking as recommended by Birrell and Ould (3). This was done, in particular, to check for common faults such as syntax errors, wrongly nested conditional statements, incorrectly used variables and uninitialised variables. After correction of any programming errors revealed by this method, each program unit was then verified by a formal, or semi-formal, use of axiomatic semantics as outlined in Backhouse (1).

This involved the specification, from the software design, using predicate calculus of pre- and post-conditions on code modules. The post-condition defines the state that the program is required to be in after the execution of that unit, whilst the pre-condition defines the state that the program will be in before execution of that unit. From the post-condition, program synthesis then works backward to establish the weakest pre-condition, i.e. a predicate describing the set of all initial states such that execution of the intermediate code begun in any one of these states is guaranteed to terminate in a state satisfying the post-condition. It is then necessary to show that the pre-condition subsumes this weakest pre-condition.

Such a thorough verification that software is built correctly is rarely reported for sports biomechanics applications.

SOFTWARE TESTING

Each program unit was tested with specimen data sets obtained from equivalence partitioning of the input and output data space, along with boundary-value analysis to exercise the module on the boundaries of the equivalence classes. This black box approach, as recommended by Ould and Unwin (8), allowed the development of representative and comprehensive test cases. It was supplemented by the use of structural (white box) testing using the independent paths coverage criterion. This ensures the execution during testing of a complete set of linearly independent paths through the code and requires a number of tests equal to the cyclomatic complexity of the module.

This testing procedure was then repeated for each complete program within the overall package. The purpose of testing in this way, which concurs with the recommendations of Myers (7) was to ensure as far as reasonably possible that the correct product was built, i.e. the validation of the software.

USE OF THE PACKAGE

After this thorough verification and validation of the software, the EMG software is being used in several studies of sports movements, e.g. an investigation of wing kayak paddling techniques, research into the bio-mechanics of archery and stress reduction in golf putting. Fuller details of the results of some of these studies will be reported elsewhere.

The software reported herein is sufficiently machine independent to be easily adapted to other computers, and should therefore be of interest to many researchers in the field.

ACKNOWLEDGEMENTS

The author wishes to acknowledge the financial support provided by the British Sports Council to enable him to attend the ISEK Symposium.

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EMGGEN: A SOFTWARE PACKAGE FOR MYOELECTRIC SIGNAL SIMULATION DESIGNED FOR RESEARCH AND COMPUTER AIDED INSTRUCTION

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INTRODUCTION

In the last decade the rehabilitation community has expressed increasing interest in surface electromyography. A number of authors [6,7,11,9,1] have shown that the surface myoelectric signal must not be considered less accurate or informative than that obtained with percutaneous methods, and that the two techniques are actually different and complementary. Among other applications, EMG spectral parameters may provide information about muscle fatigue during sustained contractions [6,9,4], and various techniques may be used to obtain the non-invasive estimation of muscle fiber conduction velocity (CV) [10,5].

Mathematical models of the power spectrum of the myoelectric signal have been proposed by several authors. Among others, Basmajian and De Luca [2] (Chap. 3) and Lindstrom [8] reported a detailed mathematical analysis of the EMG generation mechanism and a possible generation model. In order to interpret correctly the information content of surface EMG signal some fundamental concepts must be clearly understood. It is well known that the features of surface EMG depend on several factors, and namely the number of active motor units, the anatomical and physiological characteristics of each motor unit, the time course of its firing rate and CV, the superposition of the action potentials generated by different MUs, the filtering effects of tissues, and the detection electrode and EMG amplifier. Although the relationships among the EMG power spectrum and some of the factors previously mentioned have been studied both theoretically and empirically [2], the interpretation of the information content of surface myoelectric signal is still fragmentary and sometimes controversial. Moreover, for people trained in life sciences it is generally difficult and time consuming to understand satisfactorily the mathematical concepts on which EMG generation models (and the actual EMG signal) are based.

The aim of this work is to describe *EMGGEN*, a software package recently developed that may be used either as a research or a teaching tool. In fact, by simulating signals whose characteristics are determined *a priori* and correlated with physiological and anatomical correlates, *EMGGEN* allows the researchers to gain a deeper understanding of the properties of the myoelectric signal and to evaluate objectively the features of different signal processing techniques. Moreover, its interactive structure is well suited for teaching purposes, allowing the students to verify the effect of different anatomical, physiological and technical factors on the myoelectric signal characteristics, both in time and frequency domains.

DESCRIPTION OF THE PACKAGE

Based on an extended version of Lindstrom's model, *EMGGEN* simulates either internal or surface myoelectric signal collected during voluntary or electrically elicited contractions. Both monopolar and bipolar electrodes may be simulated. When a bipolar electrode is selected, the user may generate either single or double-differential signals. The first option is useful to investigate the myoelectric signal spectral properties, the latter has been implemented mainly to evaluate non-invasive estimation techniques of muscle fiber CV [3].

The software package consists of three sections: I) Interactive definition of the characteristics of the signal to be synthesized, II) generation of the signal, and III) estimation of the time course of amplitude and spectral parameters of synthesized signals.

In the first section of the package the user defines interactively the number of active motor units, their geometrical relationships with respect to the electrode (depth, distance from the end-plate zone, angle between the fibers and the skin), the initial value of muscle fiber CV and its variation modalities (linear or exponential), and the firing history of each motor unit. The signal to be generated may consist of up to 100 distinct motor units, obtained from a library of 10 different templates. These were extracted from real myoelectric signals collected by needle-electrodes and processed by a decomposition algorithm [2]. Through the friendly man-machine interface, the user may verify immediately the effect of each choice on the motor unit action potential, both in time and frequency domain. Action potentials and their power spectra may be stored temporarily and compared, thus enabling a personalized study of the effect of the variation of particular parameters. Fig. 1. represents different MUAPs and their power spectra obtained by changing the depth of the active motor unit. Similar plots may be obtained when other parameters are changed.

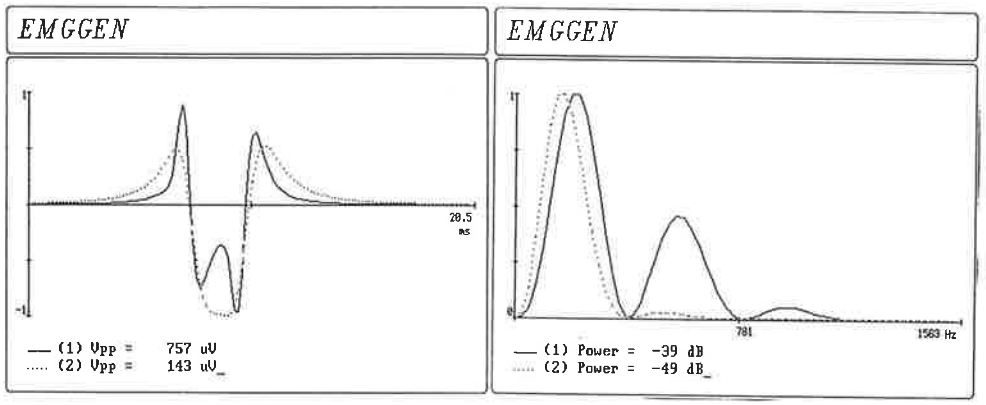

Fig. 1. Filtering effect of the tissue: (Left) Action potentials obtained for different depth of a motor unit, and (Right) Corresponding power spectra (Differential electrode, 10 mm interelectrode distance, CV = 4 m/s).

After defining the geometrical characteristics of the active motor units, the user enters the initial CV of each motor unit and the related modality of variation (linear or exponential, increasing or decreasing). Each motor unit may be caused to be recruited and derecruited in different time epochs during the simulated signal. When a motor unit is recruited its firing rate may be constant, linearly or exponentially increasing or decreasing, or may change sinusoidally. The inter-pulse interval (IPI) variation is considered as a normally distributed random variable, whose standard deviation is selected by the user. Again, the selection of the firing history is totally interactive and the user may verify immediately the effect of the choices. Fig. 2. represents the summary of the characteristics and a plot of the firing history of a selected motor unit.

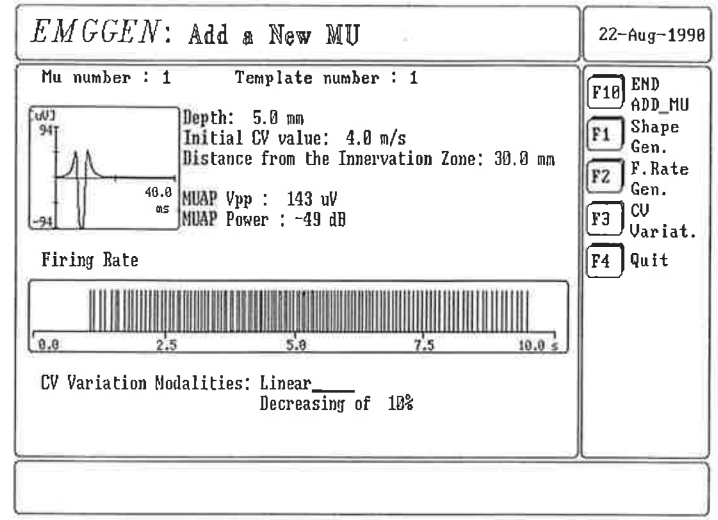

Fig. 2. Summary of the characteristics of the action potential generated by an active motor unit, and its firing history.

The second section of the package allows the user to generate the signal. The user may generate either a single-differential signal or a single-differential and two double-differential signals. Changes of CV during the simulated contraction affect both the shape of the selected templates and the filtering functions of tissues and electrodes. A signal generated with ten motor units with different anatomical characteristics, firing rate behavior and CV variations is presented in Fig. 3.

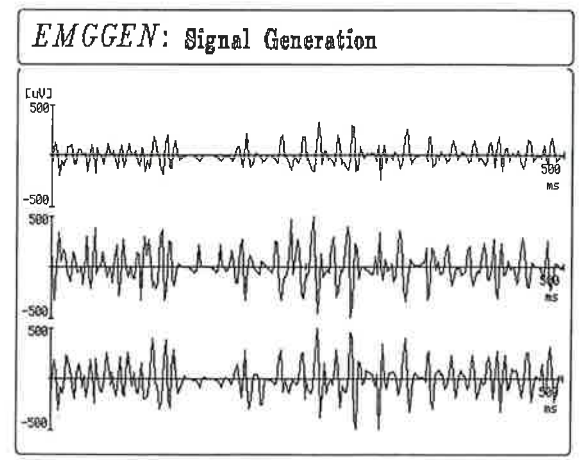

Fig. 3. Simulated signal obtained by summing the contribution of 10 different motor units.

The third section of the package enables the user to estimate spectral parameters, amplitude parameters, and CV. This section is useful for teaching purposes, because students may verify the effect of technical (interelectrode distance) or anatomical factors on the shape of the observed power spectra and CV estimates. The effect of different estimation techniques on spectral variables may also be investigated.

CONCLUSION

In conclusion, a software package running on AT-compatible computers with DOS operating system has been developed for research and teaching purposes. The man-machine interface has been designed to enhance the didactic capabilities of the package. The options included in the package allow the user to simulate numerous combinations of different anatomical, physiological and technical factors that could affect the myoelectric signal detected during voluntary or electrically elicited (isometric or dynamic) contractions. In order to take full advantage of the teaching capabilities of *EMGGEN* students must be exposed to the theoretical issues of the generation of the EMG signal before using the package, that may then be used as a tool to test the comprehension of the theory and improve it.

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PEDAGOGICAL, ERGONOMIC AND COMPETITIVE APPLICATIONS OF KINESIOLOGICAL EMG (IN ALPINE SKI)

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INTRODUCTION

Measuring muscle activity of Alpine ski movements in field conditions is amongst the most difficult kinesiological investigations because there are too many influencing factors - e.g. snow conditions - temperature - that cannot be controled.

Although many studies have examined - the skier's movements, - the role of the musculature in relation to the high potential risk of injury, - the issue of importance of the release of ski bindings and its design, - the muscle strength of skïers and the substantial anoerobic power requirements, very few studies have "filled the gab" between electrophysiological research and the practical approach of motor learning aspects, the choice of skï material and competition feedback through EMG.

The principal objectives of this alpine ski & EMG project are based on sugestions of the French ski school of Tignes - Val d'Isère where the experiments were conducted. The purpose of this study, therefore is (i) to determine the muscular activity during the 3 basic turns (- the Stem Christie, the Stem Turn and the Parallel Christie) of the ski-initiation learning process with the emphases on the "flexion-extension" mecanism of the lower limbs (the pedagogical application) (ii) to compare muscular activity using "compact", "soft" and "competition" type of skies both during turns and down hill racing on a 11° and a 27° slope (the ergonomic application), and (iii) to measure EMG during simulated special Slalom and Giant Slalom conditions and how the muscular activity is affected by different slope inclinations and snow conditions (the competitive application).

MATERIAL AND METHODS

Subjects were 31 experienced ski instructors of the French ski school (N = 25 for the study i and ii and N = 6 for study iii) and were, prior to the experiments, informed of the nature of the testing procedures.

The analog raw EMG was recorded on location with a portable seven channel FM recorder (TEAC MR30) and with preamplified bipolar surface electrodes supplied with a precision instrumentation amplifier (AD 524, Analog Devices Norwood, USA). These active electrodes were fixed on the midpoint of six lower limb muscles including agonist - antagonist muscle pairs. The hardware for the EMG data acquisition was designed for multidisciplinary purposes. The skier was not to be disturbed during the movement. The system has a freedom of action (continuous measurement over several minutes). Several muscles as well as synchronisation signals are monitored simultaneously. Influences of skin resistance phenomena were eliminated by means of high imput impedance amplifiers (Clarys-Publie, 1987). The raw EMG was full wave rectified and enveloped using a moving average principle and normalized to the highest peak amplitude procedure per subject and integrated (Winter et al., 1980). Further analyses procedures were carried out with the Electromyographic Signal Processing and Analysis System - E.S.P.A.S. (Cabri, 1989).

Qualitative pattern specificity characteristics were analysed with the IDANCO-EMG pattern evaluation system (Clarys-Cabri, 1988).

RESULTS

In the 3 studies and for all subjects (Elite skiers - N = 25 ; N = 6) we have found a high level of co-contractions of the lower limb extensors and flexors, especially during the extention phase of the ski movement (Fig. 1). The Stem Christie and the Parallel Christie showed a higher level of rhythmic movement (92% and 84%). The second initiation movement of the Stem Turn produced 74% of rhythmic repetitive patterns. In addition its agonist-antagonist coordination pattern was more complex.

The activity relation between the agonist:antagonist pairs (2 flexors versus 2 extensors) is clearly different at the level of co-contraction in the Stem Turn if compared to the 1st initiation movement (Stem Christie) and to the 3rd initiation movement (Parallel Christie). These findings might explain why ski instructors encounter initiation problems teaching the second skill (Stem Turn).

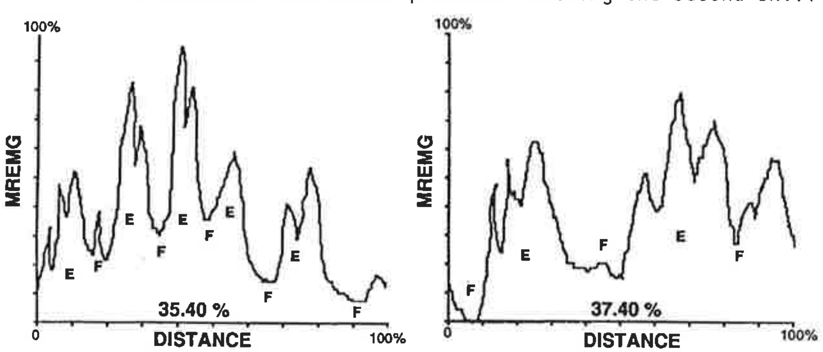

Figure 1: Mean rectified EMG (25 subjects + 6 muscles) of the flexion-extension movement on a vaulted downhill slope (left) and of the parallel christie (right)

The differences between the mean rectified EMG data of dynamic contractions while skiing with different types of skis and the mean rectified EMG data of the highest peak were used in the primary analysis of data. Based on this comparison, differences between the effects on muscle activity of the three types of skis were unimportant. In a second phase, the normalized linear envelopes of all subjects were graphically superimposed and averaged (Clarys et al., 1986). The EMG data were considered in combination with anthropometric values, with snow characteristics and with the velocity of skiing. This showed systematic differences between the use of the racing, soft and compact ski.

If one accepts that "the lower the muscular envolvement, the better the execution of movement" is done at identical velocities, the soft ski could be presented as the better overall ski for recreational skiing.

The comparison of L.E. patterns and of IEMG while through flat and excavated, but identical special slaloms (slope and distance) indicated no dignificant differences, nor at the level of velocity (e.g. time) to go through the slalom. Analysis per muscle and per subject and all data combined showed no differences either (Fig. 2). But verification of the IEMG over the slope inchination (and heartrate) during a similated (but short) Giant Slalom indicates that the muscular activity decreases and increases with the slope angle, while e.g. heartrate constantly increases (Fig. 3).

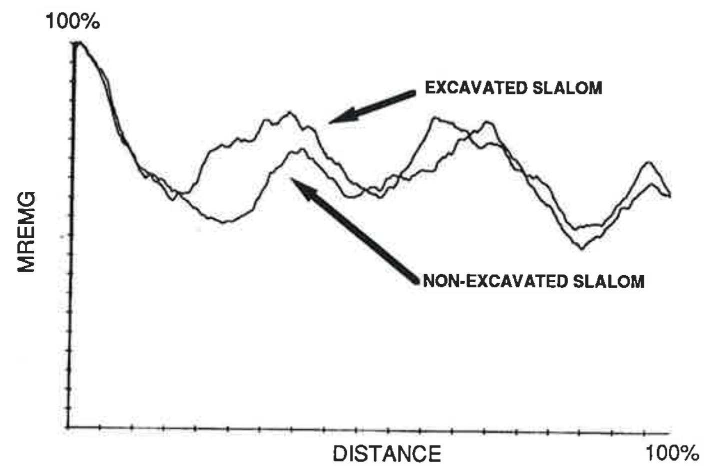

Figure 2 : Mean rectified EMG (6 subjects + 6 muscles) on an excavated and a non-excavated Special Slalom

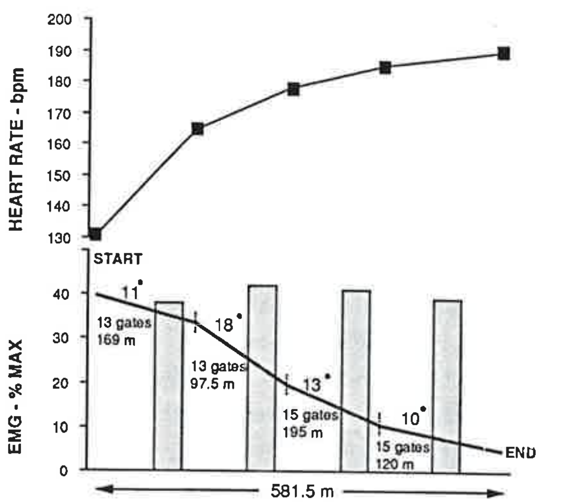

Figure 3 : Muscular intensity, slope inclinations + distance + no. of gates, and heart rate during (part of) a Giant Slalom

CONCLUSIONS (Feedback)

These data suggests that (i) the Stem Turn demands a higher level of neuro-muscular ability than the other two learning drills and it should therefore be introduced in the learning process of more advanced skiërs; (ii) on average and for the slope inclinations investigated (11° and 27°) the soft ski showed the least EMG activity so that the idea that it allows better control of the overall ski movements can be supported ; (iii) the excavated snow conditions in slalom competitions do not disadvantage the athletes.

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EMG INVESTIGATION OF THE EFFECTS OF PERIPHERAL FEEDBACK ON GOAL DIRECTED WRIST MOVEMENTS

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INTRODUCTION

Since 1987, the Orthopaedic Biomechanics Laboratory at Shriners Hospital in San Francisco has been investigating the phasic relationships of the hand and thumb musculature during functional tasks. Earlier studies demonstrated that the normal patterns of muscle activation were characterized by individual differences that were consistent and predictable for each subject. We proposed that subjects utilize a "motor planning strategy" incorporating gravity and tenodesis effects, in combination with muscle action, to accomplish specific motor goals. The current study was to examine an individual's motor planning strategy when the motor goal or sensory feedback is changed.

METHODS

Five normal adult volunteers were studied. Fine wire electrodes were inserted into extensors carpi radialis longus and brevis, extensor digitorum communis, extensor carpi ulnaris, flexor carpi radialis, flexor digitorum superficialis, and flexor carpi ulnaris of the dominant arm. The EMG signals were differentially amplified and cabled to a computer. Electrogoniometers were used for recording wrist and finger motion during the performance of the test activities. The angular velocity of each movement was calculated as a derivative of the angular data from the goniometers and was included in the printouts of the data.

The subjects performed a series of wrist flexion and extension motions with variations in both the motor goal and the sensory feedback. Five arcs of motion were tested, including maximum wrist flexion from neutral and from full extension, maximum wrist extension from neutral and from full flexion, and cyclic wrist flexion and extension.

Tests were performed under two different conditions which we termed "targeted" and "constant feedback". For the targeted tests the subjects were instructed to flex and extend the wrist through the series of motions as quickly as possible, without visual feedback, until a physical stop terminated the movement. Arcs of wrist flexion and extension were limited by adjustable pegs attached to a smooth board. By changing the position of the pegs, different arcs of motion were tested.

The "constant feedback" motions required the subjects to control the speed and position of the wrist by following a set trajectory path as accurately as possible. For this test, the wrist goniometer output was displayed by an oscilloscope trace. The subjects were asked to use the oscilloscope trace to follow a line drawn on acetate which was placed in front of the oscilloscope screen.

Vibration of 160 Hz. was applied over the proximal muscle bellies of the extensors of the wrist for 60 seconds or until a tonic vibration reflex was observed. Cyclic flexion and extension were repeated under both the targeted and constant feedback conditions.

A topical anesthetic with 20% Benzocaine was sprayed or rubbed on the skin over the wrist flexors for 5 to 10 seconds. Cyclic flexion and extension was again repeated under both targeted and constant feedback conditions. EMG output was recorded 5 minutes after application of the spray.

Motion of the wrist was used to identify the specific phases of each task. Flexion and extension phases were based on changes in wrist position and a sharp drop in angular velocity indicated when the subject hit the physical stop during the targeted tests. Muscles were classified as agonists or antagonists during each of the test activities. The time of onset, relative to the beginning of the movement, and the duration of EMG activity were described.

RESULTS

In targeted wrist movement, fast wrist flexion or extension against a stop activated all of the agonist muscles sampled in this study for all five subjects. The onset of activity was always at the initiation of movement and continued until hitting the stop for most subjects. The EMG amplitude was characteristically large relative to the amplitude of EMG from the same muscle in other test activities. Relatively lower amplitude activity was observed in one or two, but not all of the antagonist muscles during the fast targeted movements.

Individual subjects were very consistent in which antagonist

muscle was active, but there was no consistent pattern across subjects. ECU was used as an antagonist to wrist flexion by three subjects and two subjects used ECRB and EDC as antagonists consistently during wrist flexion. In one subject, no antagonist activity was recorded.

The wrist flexors were generally less active as antagonists and were not active at the onset of movement for most subjects. Antagonist activity during extension was recorded consistently from FCU for two subjects, and one subject activated FDS during wrist extension.

Under the constant feedback conditions, four of the subjects activated all of the agonist muscles during flexion or extension of the wrist. The one subject who was the exception did not use FCU during wrist flexion or ECU during wrist extension. Individual differences were observed, but were consistent for each subject. Under the constant feedback conditions, agonists that were active at the onset of the movement during the targeted movements, were more active near the end of the range under the constant feedback conditions. Subjects increased and decreased EMG amplitude during the feedback motions to control the wrist position. More co-contraction of agonist and antagonists was observed when the subject performed wrist movements using constant visual feedback.
During wrist flexion, all five subjects recruited at least one more antagonist muscle than with wrist flexion performed under the targeted condition. Four of the subjects recruited two additional antagonist muscles. ECRL was used by all five subjects in at least one run of wrist flexion. In contrast, ECRL was not active during targeted wrist flexion.

Patterns of activation and duration of EMG activity were essentially unchanged by the vibration or anesthetic spray.

DISCUSSION

The results of this study indicate the timing of muscle activity and the activation of agonist and antagonist muscle groups is different when the goal of movement is changed and different sensory feedback is required. During the targeted movements, the subjects relied on peripheral feedback from hitting the stop to signal the end of agonist activity and the activation of opposing muscle groups to perform the reciprocal motion. Under conditions of constant feedback, a balance of agonist and antagonist activity was needed to give precise control of the motion. To accomplish this, subjects varied both the amplitude and the activation times of agonist and antagonist groups.

ECRL was not used as an antagonist during targeted wrist flexion, but was frequently used as an antagonist during wrist flexion with constant feedback. This muscle may function more as a postural or balancing muscle than a primary motor force in controlling the wrist.

ECU was active to assist in the final degrees of wrist flexion as reported in previous studies. We observed this in some of the tests as well. In addition, for one of the subjects, the timing of ECU during wrist flexion was actually synchronous with the wrist flexors, active at the onset of movement. This unusual timing pattern was only observed in one subject, but it suggests ECU may play a more important role during wrist flexion than has been previously appreciated.

In the present study, no changes could be observed with either the vibration of the extensor muscles or with application of an anesthetic spray to the volar surface of the forearm. While no changes in muscle timing could be demonstrated in normal adults, it is possible that vibration and/or cutaneous anesthesia may alter muscle timing in patients with central nervous system disorders.

The data in this investigation support the overall concept of motor planning strategies. Selection of both agonist and antagonist muscles is consistent for individuals, but not universal. The way that a muscle is used in the performance of a specific task appears to depend on the motor strategy of the individual. Changes in the sensory feedback required to perform the motor task result in modifications in the motor strategies. However, the selection of specific agonist and antagonists remains a characteristic of individual normal subjects.

Better knowledge of the role of sensory input and variations in motor goals can improve understanding of the neurologic control of upper extremity muscles. Such understanding is important for the design and evaluation of treatment modalities for patients with upper extremity dysfunction. This information may also be useful for planning protocols for dynamic EMG testing in patients.

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PERIPHERAL INDUCTION IN GENERATING ELEVATOR MUSCLE ACTIVITY DURING SIMULATED CHEWING IN MAN

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INTRODUCTION.

The chewing rhythm is fairly constant in human mastication (1). To maintain a constant chewing rate the activity of the masticatory muscles has to adapt quickly to the resistance of food. Without food, a low level of muscle activity has been observed in the surface EMG of subjects making pseudo chewing movements (2). In contrast, a large Additional Muscle Activity (AMA) occurred to overcome a resistance if food was present. The aim of this research was to examine how the AMA is controlled by the nervous system.

Two hypotheses can be stated for the control of the AMA. First the AMA may be generated by a Central Pattern Generator (CPG). A CPG is a neural network which programs muscle activity, i.e. its onset, duration and amplitude. A pure CPG for the control of the AMA is unlikely, because of variation in consecutive chewing cycles of the size of the food bolus, its particle size composition and the location on the tooth arch where the first engagement of the bolus occurs between antagonistic teeth. If a CPG controls the AMA, it will probably be reset each chewing cycle by using peripheral information about the food, so that an anticipating character of the AMA can be expected.

Second, the AMA may be peripherally induced. Peripheral stimuli may originate from pressure receptors in the periodontium, from muscle spindles or from receptors in the Temporal Mandibular Joint. If the AMA is peripherally controlled, events occurring in the actual chewing cycle determine the muscle activity. In that case, the AMA will not show an anticipating character, since in every chewing cycle it depends upon the engagement of the food between antagonistic teeth.

A direct way to investigate the origin of the AMA would be to make a subject chew on food and remove the food unexpectedly out of the mouth. Or, reversely, to place food unexpectedly into the mouth while the subject makes chewing movements. The adaptation of the muscle activity to the changed circumstances will give insight in the control of the AMA. Although such food manipulation is not possible during normal chewing, these situations can be approximated during rhythmic open-close jaw movements, where food resistance is simulated by a computer controlled magnet-coil system.

METHODS AND MATERIALS.

Nine subjects, participating in the experiments, made rhythmic open-close movements at their natural chewing rate, controlled by a metronome. A coil was located in a permanent magnetic field and rigidly attached to the mandible by means of a clutch, cemented upon the teeth. The mandible could be loaded in a downward direction by means of a computer controlled electric current in the coil. The jaw gape was measured by means of an optical motion analysis system (Selspot ^{R}). The force started at a fixed value of the jaw gape during the closing phase of the mandibular movement. The amplitude of the force increased proportionally to the decrease in jaw gape up to a maximum of 24 N. After the mandible had reached occlusion, the force smoothly disappeared in within 200 ms. EMG was recorded from the masseter muscle and the anterior temporal muscle on both sides by means of bipolar surface electrodes. The EMG signals were amplified, filtered (bandpass 10 - 1000 Hz) and further rectified and smoothed. The EMG signals, the force signal and the position signal were all sampled with 500 Hz.

Two types of experiments were performed. The first type is denoted as DISAPPEAR experiments, because the force disappeared at a random cycle after a large number of open-close movements with a force. The last two cycles with force and at least eight cycles without force were recorded. The second type is denoted as APPEAR experiments, because at a random cycle - after a large number of open-close movements without force - the force appeared. The last two cycles without force and at least eight cycles with force were recorded. All measurements were repeated at least 30 times and averaged to reduce experimental scatter. Furthermore, the averaged EMG activity was corrected for the activity required to make open-close movements without activating the magnet-coil system.

RESULTS.

Fig. 1 shows a representative example of a DISAPPEAR experiment. In cycle 1 and 2, with force, a large AMA is present. In cycle 3, the first cycle without force, the AMA has almost completely disappeared, in particular that part of the AMA occurring 30 ms and later after the onset of the force. Since this part of the AMA only occurred when the force was present it is obviously not elicited by an anticipating mechanism in which a CPG may be involved, as the subject could not expect the disappearance of the force. This large contribution to the AMA is therefore peripherally induced , the more since it started after the onset of force. However, a small contribution to the AMA, starting about 70 ms before the onset of the force, was still present in cycle 3. In the following cycles this contribution had also disappeared. Since this small contribution started before the onset of the force when a force was expected, this contribution will further be denoted as the anticipating AMA contribution.

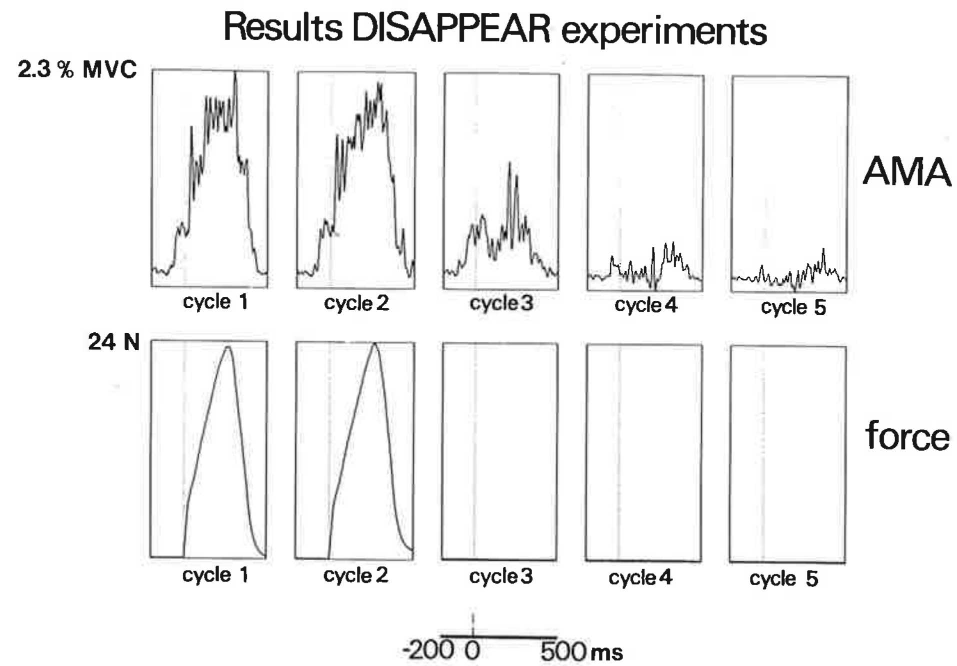

Fig. 1: Mandibular load and averaged Additional Muscle Activity (AMA) in the right masseter muscle during a DISAPPEAR experiment.
Top row : The averaged AMA. (Rectified and smoothed).
Bottom row : The force applied on the mandible during jaw-closing.
The dotted vertical line in each chewing cycle indicates the moment at which a force could be evoked. EMG recordings were averaged over an interval of -200 to +500 ms with respect to this point.
The activity in cycle 3, about 170 ms after the onset of the force, was not part of the AMA, but was related to clenching in occlusion.

Fig. 2 shows a representative example of an APPEAR experiment. In cycle 1 and 2, when no force was present, no AMA was observed. In cycle 3, the first cycle with a force, a large AMA was present. The AMA started late, 80 to 160 ms after the onset of the force. In cycle 4, the second cycle with force, the large AMA component which depends upon the presence of the force (See fig. 1, DISAPPEAR experiments) started earlier after the onset of the force, i.e. after about 30 ms. The small anticipating contribution was also observed, starting about 70 ms before the force.

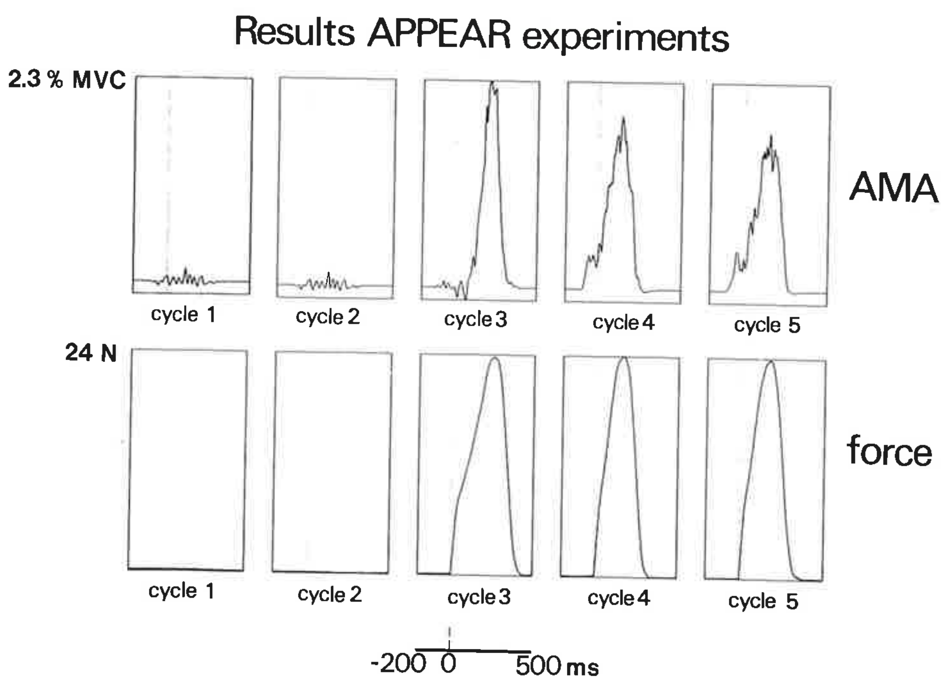

Fig. 2: Mandibular load and averaged Additional Muscle Activity (AMA) in the right masseter muscle during a APPEAR experiment.
Fur further explanation see Fig. 1.

Whereas the DISAPPEAR experiments show that most of the AMA is peripherally induced, the APPEAR experiments demonstrate that this AMA contribution does not have a constant latency. The latency was long (about 120 ms) in the first cycle with force, but short (about 30 ms) in the following cycles with force, indicating that the neural mechanism controlling the AMA quickly adapts to the force, i.e. within one cycle. Hence, a mandibular load may initiate a neural process of preprogramming the AMA, which can only be released by a peripheral trigger in the subsequent cycle.

The small anticipating contribution to the AMA observed in both types of experiments may be under control of a Central Pattern Generator and may also be the consequence of applying the force each time at the same jaw gape.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

EFFECT OF MOVEMENT FREQUENCY ON EMG AND LEG FORCES DURING DYNAMIC EXERCISE IN HUMANS

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INTRODUCTION

The concept of optimum movement frequencies has attracted the attention of many physiologists dating back to A.V. Hill in 1922 [1] and beyond. In the present study we have examined the effect of varying movement frequency on the force required and the EMG elicited, during progressive dynamic exercise performed pedalling a cycle ergometer.

MATERIAL AND METHODS

Four healthy male subjects came to the laboratory on 3 occasions to exercise at either 40, 70 or 100 rev/min. On arrival at the laboratory they first performed a maximal sprint effort of a few seconds duration at the chosen constant pedalling rate. This was done using the isokinetic cycle ergometer previously described [2]. The cranks of the ergometer were instrumented to allow continuous recording of the 'effective' forces exerted across them. However analysis in this paper is confined to the peak force - that is the greatest force exerted by one leg during each revolution (see [2,3,4]). Maximal peak force exerted on the cranks was recorded along with the EMG from selected knee extensor muscles using surface electrodes with built in pre-amplifiers (Medelec EA 1000). The signals were amplified and full wave rectified and an integration of the signal made every 100ms (IEMG).

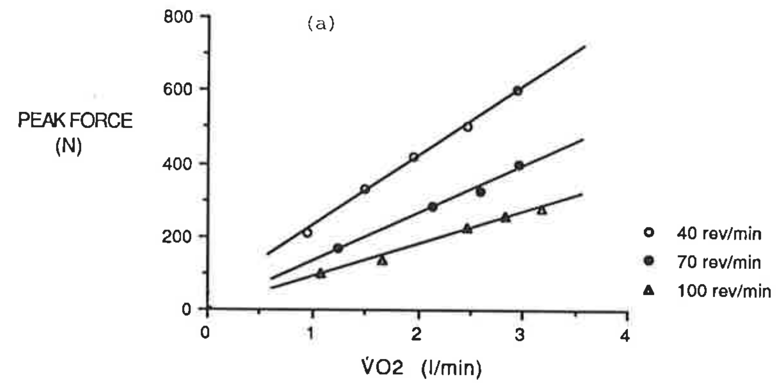
After resting, with the EMG electrodes undisturbed the subjects performed a progressive multi-stage exercise test on the same cycle ergometer but with the isokinetic system disconnected and the chain wheel connected to a conventional friction braked ergometer. During the last minute of each 5 minute stage oxygen uptake, forces on the cranks and EMG were recorded. The peak force exerted was measured in 30 consecutive revolutions and averaged (mean peak force).

Statistical comparisons were made using Student's paired t-test. (Mean and SEM data are reported)

RESULTS

As shown for one representative subject mean peak force increased linearly with exercise intensity at each pedalling rate. Further at any given exercise intensity the peak force required decreased as pedalling rate increased (Fig 1a).

However the maximum peak force available also decreased as pedalling rate increased in accordance with the force/velocity relationship (see also [2]). Nevertheless even when the peak force was expressed as % of the maximum available force (i.e. measured at the same velocity) the differences between the experiments persisted (Fig 1b).



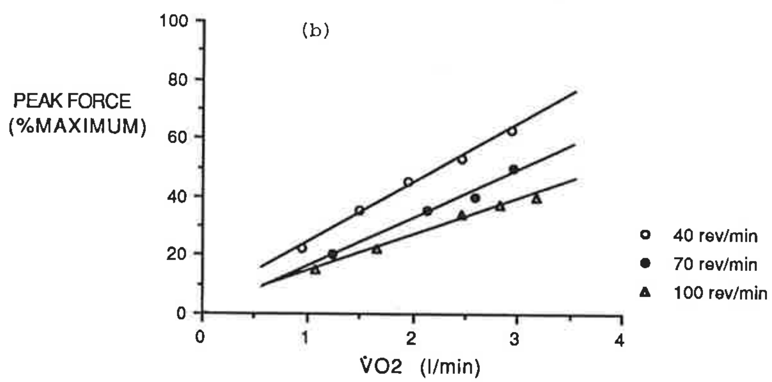


Fig. 1 Mean Peak Force in (a) absolute units (N) and (b) expressed as % of maximum available force. Both plotted in relation to exercise intensity expressed as oxygen uptake ($\dot{V}O_2$) 3 experiments at 40, 70 and 100 rev/min - subject No.1.

Taking the data for all 4 subjects the proportion of maximum available force utilised when exercising at an exercise intensity equivalent to 75% $\dot{V}O_2$max fell from 58 ± 6% at 40 rev/min to 44 ± 6% at 100 rev/min (Fig 2: $p < 0.001$).

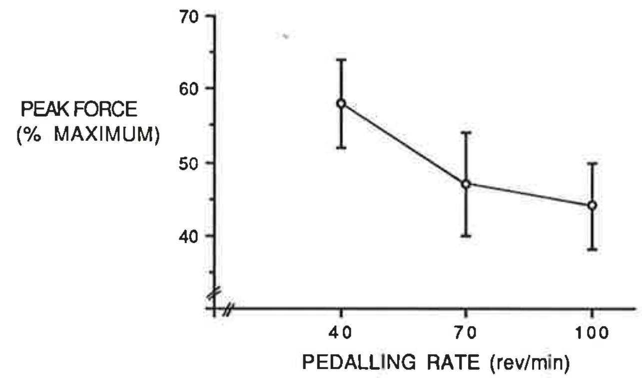


Fig 2. Proportion of Maximum Peak Force utilised at 75% $\dot{V}O_2$max performed pedalling at 40, 70 and 100 rev/min (Mean ± SEM of 4 subjects)

This decrease in the proportion of maximum available force utilised was also reflected in the IEMG when expressed as % of the level recorded during the maximal sprint at the same velocity. Thus, typically for vastus lateralis muscle % maximal IEMG at 75% $\dot{V}O_2$max was 51 ± 6, 42 ± 4, and 40 ± 8% at 40, 70 and 100 rev/min respectively (Mean ± SEM of 4 subjects).

DISCUSSION

These results indicate that at an exercise intensity equivalent to 75% $\dot{V}O_2$max there was a greater reserve of force generating capacity at high compared to low pedalling rates (see also [5]). This greater reserve is also reflected in the EMG. This greater 'reserve' suggests a lower peripheral (muscle) stress. Thus this could be an important strategy for delaying the onset of fatigue during prolonged high intensity exercise.

The general pattern of the relationship between proportional force utilised and pedal rate as illustrated in Fig. 2 has been found in all subjects that we have now measured (n > 50). However it must be realised that the level of the relationship will reflect the balance, within an individual, between aerobic function and muscle size strength and power. In another study we found, for example, that at an exercise intensity of 75% $\dot{V}O_2max$ (pedal rate 70 rev/min) international weightlifters used only 25-30% of their available force while endurance athletes used 50-60%. The latter value representing the very high aerobic quality of a relatively small muscle mass.

Furthermore the possible benefit of choosing a fast rather than slow pedal rate will depend on the relative fatigue resistance of the population of muscle fibres recruited. This may be an especially important consideration at higher exercise intensities where a large part of the fibre hierarchy is already recruited (6).

These findings have relevance to both athletic performance and clinical practice. Firstly they may explain the choice of apparently fast pedalling rates by competitive cyclists in endurance events. For example the mean pedalling rate during the world record for the 1 hour unpaced event (1956-1972) has been reported as 105-108 rev/min. It has been reported that such high pedal rates incur an additional oxygen cost for a given power output (i.e. reduces gross mechanical efficiency) which would be a disadvantage when athletes are already operating at or close to their $\dot{V}O_2max$. However we have only found this effect at relatively low exercise intensities. In trained endurance athletes exercising at 300W power output we could detect no differences in gross efficiency between experiments performed at 40, 70 or 100 rev/min (7).

Finally these observations have implications for exercise testing in the laboratory and in clinical settings. Subjects or patients with muscles which are relatively small or weak, in relation to their aerobic function, may find it impossible to achieve maximum oxygen uptake at the slower pedalling rates which are often recommended.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

MUSCULAR RESPONSE TO SUDDEN ANKLE INVERSION - DIFFERENCES BETWEEN A TAPED, BRACED AND BARE ANKLE

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INTRODUCTION

The ankle joint is the most frequently injured joint of the human body in sports (1). Mostly the injury is due to an inversion movement. During the ankle inversion muscle spindles and joint mechano receptors are activated. Glencross and Thornthon (2) showed a reduced joint position sense in subjects who had an ankle sprain. This reduced propriocepsis increases the risk for recurrent ankle sprains.

The muscular apparatus to prevent inversion is relatively small. The peroneus muscle is the most important muscle involved in the counteraction of the inversion. Tropp et al. (3) showed that the pronator muscles were weaker on the affected side in unilateral functional instability of the ankle joint. Strengthening of these muscles may aid in the prevention of ankle injuries.

To prevent the occurrence of an ankle trauma, several kinds of supports are used, such as tapes and braces. The theory behind supporting an ankle is, that the support will take over the high forces on the lateral side of the vulnerable ankle during inversion movements. However, this 'mechanical' support changes in time; Rarick et al. (4) showed that the mechanical support of tape decreased with 40 percent after ten minutes of exercise.

It is possible that tape and other supporting devices not only give a mechanical support, but they also stimulate sensors in the skin, which make the pronator muscles contract sooner and/or stronger. Glick et al. (5) concluded from running experiments that "taping had an action stimulating the peroneus brevis muscle".

These effects of ankle support, as proposed in the literature, were investigated in this study. If the supports have a mechanical effect, the pronator muscles need to generate less force. This should be visible as a reduced electromyographic response of these muscles. Further, if a reflex mechanism is involved, the pronator muscles must contract in an earlier stage after the inversion. Moreover, the electromyographic response may be increased.

To test these hypotheses the electromyographic response of the peroneus longus muscle was recorded in three experimental conditions: when the ankle was taped, braced or bare. The tibialis anterior muscle was also recorded because this muscle can counteract the plantar flexion, which is very often part of a severe ankle sprain.

MATERIALS AND METHODS

Nine healthy subjects (3 male, 6 female, aged 20 to 24 years) without previous ankle trauma participated in the study. Every subject underwent 5 consecutive ankle inversions in three conditions: with tape, with a brace and barefoot. A balanced latin square design was used to negate order effects of the experimental conditions.

The tape was 3.75 cm wide (Hansa, Bayersdorf) of heavy quality and was applied in 3 supporting bands under the foot to 20 cm above the malleoli. Two transversal bands were applied for fixation. The brace was a push-brace of medium strength, often used in the Netherlands, which consisted of an ankle tube fixed with elastic and non-elastic bands. These braces are available in six sizes, three of which were used in our investigation.

The subjects were standing on a platform, with the left foot on a balance and the right foot on a trap door. The trap door opened 30 degrees around the sagittal axis and 8 degrees around the frontal axis, because the ankle inversion is a movement around more than one anatomical axis. The trap door was opened at random time intervals, but only as the weight on the left foot was about half the body weight.

Bipolar surface electrodes registered the muscular response. They were applied on the belly of the peroneus longus and tibialis anterior muscle. The reference electrode was placed at the lateral femoral condyle. The skin was sanded and cleaned with alcohol before application of the electrodes. The electrodes were tightly fixed with elastic bands.

The electromyographic signals were low-pass filtered at 500 Hz with 48 dB/octave. The signal of the tibialis muscle was high-pass filtered at 3 Hz with 6 dB/octave. The electromyographic signal of the peroneus muscle was prone to artifacts, probably caused by movement of skin over muscles. Previous investigations showed that these artifacts mainly occurred at frequencies below 30 Hz. Therefore, the signal of the peroneus muscle was high-pass filtered at 30 Hz with 48 dB/octave.

The electromyographic response to the ankle inversion was rectified and smoothed. From this response three parameters were deduced:

- the maximal peak,
- the time between opening of the trap door and the moment 60 % of the peak was reached (this parameter is called reaction time),
- the average EMG-response of 0.9 seconds after the opening of the trap door.

Analysis of variance was used to test differences between the experimental conditions. Subjects were treated as a random factor, the experimental conditions and the consecutive inversions as a fixed factor. A 5 percent significance level was used.

RESULTS AND DISCUSSION

In table I the values of the parameters for the peroneus muscle are shown for each experimental condition. In table II the parameters are shown for the tibialis muscle. For all investigated parameters analysis of variance revealed no significant differences between the experimental conditions for both muscles. Although some differences in experimental setup occur, these findings are in general agreement with the data from Sprigings et al. (6) and Nawoczenski et al. (7). Intra-subject variability was small.

Comparing the reaction time of both investigated muscles, it was found that the reaction time of the tibialis muscle was 10.7 ms shorter than the reaction time of the peroneus muscle (paired t-test; $P < 0.001$). This finding may be related to anatomical properties of both muscles.

The peak EMG decreased during consecutive inversions for the tibialis muscle. For the peroneus muscle the decrease was not significant, but an overall decrease in peak voltage was found in the consecutive experimental conditions.

The electromyographic activity before the opening of the trap door was always very low and did not change in the course of the experiment.

TABLE I
Peak EMG, reaction time and average EMG response of the peroneus muscle to a sudden ankle inversion with a bare ankle, taped ankle and braced ankle. The results are averaged over 9 subjects. No parameter showed significant differences between the experimental conditions (ANOVA, $P > 0.05$).

	bare	tape	brace
peak EMG (uV)	214	232	233
reaction time (ms)	100	98	100
average EMG (uV)	41	41	47

TABLE II
Peak EMG, reaction time and average EMG response of the tibialis muscle to a sudden ankle inversion with a bare ankle, taped ankle and braced ankle. The results are averaged over 9 subjects. No parameter showed significant differences between the experimental conditions (ANOVA, $P > 0.05$).

	bare	tape	brace
peak EMG (uV)	294	292	251
reaction time (ms)	88	88	90
average EMG (uV)	32	33	25

CONCLUSIONS

It can be concluded that an ankle support did not reduce the reaction time of the peroneus and tibialis muscle to a sudden ankle inversion. Moreover, there was no reduction in muscle activity after an inversion movement when the ankle was supported by a brace or tape.

This means that the mechanical support does not lead to a decreased muscle activity. Neither can it be concluded from our results that a reflex mechanism is involved.

ACKNOWLEDGEMENTS

The authors are indebted to J.G. Kersseboom and J.L. Noordanus (physiotherapists) and A.H.M. van Knippenberg (statistician).

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

EMG ACTIVITY WHILE PERFORMING THE ANTI-G STRAINING MANEUVER DURING HIGH SUSTAINED GRAVITATIONAL STRESS

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INTRODUCTION

Pilots of high performance aircraft are frequently required to withstand periods of high, sustained, gravitational stress (+Gz). It is imperative in these situations that the pilot maintain complete control of the aircraft while tolerating the resulting physical and physiological demands. The Anti-G Straining Maneuver (AGSM) is one of the principle methods used to tolerate these demands (1-3). Although the AGSM has been used since the mid 1940's research related to the specific muscles active, their level of activity, and contribution to an individual's +Gz-tolerance is scarce. The purpose of this study was to determine the amplitude and pattern of activity in selected muscles of the trunk and lower extremity during performance of the AGSM.

METHODS

Ten, healthy males between the age of 20 and 35 participated in this study. All Subjects were members of the acceleration subject panel at the USAF School of Aerospace Medicine, Brooks AFB, San Antonio, Texas and gave informed concent prior to inclusion in the study. Each subject was experienced with respect to riding a centrifuge and was medically cleared for exposure to +Gz.

Subject's were exposed to high, sustained acceleration using a human centrifuge. The gondola on the centrifuge was equipped with a 13 degree reclining seat, foot pedals and a standard light bar situated directly in front of the subject.

Each acceleration exposure was conducted on three different days with at least 48 hours separating each ride. Prior to each test, subjects underwent a "warm-up" ride consisting of +3Gz for 15 seconds. After a 10 minute rest period, subjects were exposed to an acceleration force of +6Gz (onset rate of +4Gz/sec.) until exhaustion (1). For all acceleration exposures, subjects wore pressurized Anti-G trousers with comfort zippers secured and bladders inflated at a schedule of 1.5 psi per +Gz, starting at +2Gz. In addition, subjects performed a maximal AGSM to avoid loss of consciousness. Subjects were able to terminate the ride at any time by releasing a manual brake. The stopping criteria used by each subject was 100% peripheral and 50% central light loss.

To determine the level of muscle activity during the AGSM, EMG activity was recorded using 3 mm diameter Ag-AgCl surface electrodes attached to the skin overlying the erector spinae (ES), external oblique (EO), bicep femoris (BF), vastus lateralis (VL) and lateral gastrocnemius (LG) muscles of the subject's dominant limb. The inter-electrode distance for each electrode pair was 2 to 2.5 cm. and a skin-electrode impedance below 10 K ohms. The EMG signals were differentially amplified, band pass filtered at 3 and 500 Hz, and then stored on magnetic tape for analysis. The EMG signals were later digitized off-line at a rate of 1024 Hz for final processing.

The amplitude of the EMG signal from each muscle was determined using the root mean square algorithm (RMS). The amplitude was then normalized to the RMS value during the first second at +6Gz. In addition, the mean power frequency (MPF) was calculated for each muscle. The MPF and normalized RMS value for each muscle was determined at one second intervals during the entire acceleration exposure. The value of each variable at 25 percent intervals during the +Gz exposure for each trial was then averaged. Because the distribution of muscle activity was found to be significantly skewed, the data were transformed using the base 10 logarithm of each value prior to the performance of any statistical tests. A repeated measures analysis of variance (ANOVA) test was used to determine if differences existed between muscles as well as during the course of the +Gz exposure. An alpha level of .05 was used for all statistical tests of significance.

RESULTS

The mean +Gz-tolerance time for these subjects was 101.2 seconds. The normalized RMS amplitude in the BF, VL and LG muscles decreased during the +Gz (Fig. 1). The ES and EO muscles, on the other hand, show a more variable pattern of activity. The results of the ANOVA revealed a significant difference in EMG activity over time as well as a time by muscle interaction. Post hoc analysis of the data indicated that the amplitude of the lower extremity muscles decreased 61.45% while those of the back and abdomen increased slightly by 3.45%. No significant difference was found between the muscles or over time with respect to MPF.

DISCUSSION

Our results show that during exposure to high, sustained, +Gz, EMG amplitude in selected muscles of the lower extremity decrease indicating that motor unit recruitment declines during the time of +Gz exposure. Based upon the MPF values, none of the muscles show signs of muscle fatigue (4). These findings indicate that performance of the AGSM does not result in significant fatigue. This finding is contrary to previous published reports (2,5). It is possible

that the well established relationship between EMG amplitude and frequency and muscle fatigue does not hold under conditions of sustained +Gz. If this is true, a different indicator of muscle fatigue needs to be used for future studies in this area.

The significant interaction found between the muscles and time illustrates the marked contrast in EMG activity observed between the muscles of the trunk and lower extremities. This pattern points to the conclusion that the muscles of the trunk are less important compared to those of the lower extremity as far as performing the AGSM and tolerating +Gz. This is contrary to previous literature indicating that the abdominal muscles are the most important (6). Further support for the importance of lower extremity muscles compared to the trunk is seen when EMG activity of the four "best" and four "worst" centrifuge riders are compared. The "Good" centrifuge riders had a mean +Gz-tolerance time of 175.6 seconds while the "Poor" riders had a mean time of 17.3 seconds. RMS activity for each group shows that the "Good" riders are able to maintain a higher level of muscle activity during the course of the +Gz exposure compared to the "Poor" riders, whose amplitude showed a steady, decrease. This is particularly true for the muscles of the lower extremity. It therefore appears that subjects who are able to maintain a higher level of muscle activity are more tolerant to +Gz.

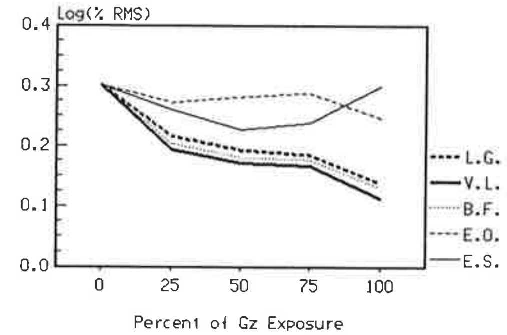

Fig. 1. Normalized RMS values for selected muscles during exposure to Gravitational Stress.

CONCLUSION

The following conclusions can be made regarding muscle activity during +Gz, 1) Motor unit recruitment decreases in the muscles of the lower extremity during exposure to high, sustained, +Gz; 2) The muscles studied, show no signs of fatigue; 3) It appears that a person's ability to tolerate exposure to high, sustained, +Gz is dependent upon their ability to maintain adequate motor unit recruitment during performance of the AGSM; and 4) Lower extremity muscles appear to have the greatest influence on +Gz-tolerance compared to those of the trunk.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

CONVERSION OF FOREARM SURFACE EMG INTO FORCE - EXPERIMENTAL DESIGN AND PILOT STUDY

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INTRODUCTION

In order to evaluate the stresses incurred during keyboard operation, the physical workload must be analyzed and forces calculated. The difficulties of quantitative assessment are related to limitations in evaluating muscular contraction in terms of force and the complexity of muscular action; particularly that of the upper extremities. An accurate analysis requires understanding of functional anatomy and biomechanics of the forearm and hand muscles. Examining the finger muscle stresses during task performance such as typing and data entry provides important information about the cause of cumulative trauma disorders. The most prevalent cumulative trauma disorder in American Society is carpal tunnel syndrome, a condition which represents increasing morbidity in the workforce. There have been several approaches used to quantify workload and force. Electromyography (EMG), especially surface EMG (sEMG), has been the most acceptable method. Surface EMG is non-invasive and easy to conduct, and has been broadly used in clinical diagnosis. Previous studies have demonstrated a linear or more than linear correlation between sEMG and the muscular forces generated (1,2,3,4,5). Technical limitations (no commercial equipment available) have not allowed the conversion of sEMG to force in finger muscles hence the need for this study. The purpose of this paper is to investigate the relationship between sEMG and force in terms of their regression equations. Flexor and extensor digitorum superficialis were selected in this study. Both muscles play important roles in sophisticated finger movements.

MATERIALS AND METHOD

Sample: Seven healthy volunteers (six females and one male;

age:22 to 50 years) participated in this experiment.

Apparatus: The Kistler 3-D force plate and the ARIEL multi-channel sEMG recorder were used. A sampling rate of 1000 Hz was selected to satisfy the Naquist criterion because the power spectra of sEMG lie between 0 - 400 Hz for both flexor and extensor digitorum. The sampling period was two seconds for each trial. The sEMG signals and isometric muscular contractions of either flexor or extensor digitorum were recorded synchronously.

Procedure: The flexor digitorum was tested with the forearm in pronation and the extensor digitorum in supination. In pronation, the subjects pressed down on the force plate with the pulp of the flexed middle finger. In supination, pressure was applied with the nail of the fully extended middle finger. In both settings, the wrist was stabilized at the external edge of the force plate. Subjects were asked to randomly apply maximal, sub-maximal, moderate and light pressure, lasting 2 seconds. Ten trials per muscle were recorded with at least one minute rest period between each pressure. The rest period served to limit the effect of fatigue.

DATA ANALYSIS

Following each experiment, the sEMG and force signals were separately processed by Analog Module of ARIEL system. The sEMG signal from either flexor or extensor digitorum was rectified and averaged by the time duration of muscle activity. The 3-D force from each muscle was recorded with three force components denoted as Fx(left-right), Fy(fore-and-aft) and Fz(vertical); and then averaged each force component on the same time duration. Since Fx and Fy were far less than Fz, the resultant force F comes as follows :

$$F = \sqrt{Fx^2 + Fy^2 + Fz^2} = Fz$$

The simple linear regression analysis was applied to all pairs of sEMG and muscle force for flexor and extensor digitorum, respectively.

RESULTS and DISCUSSION

All regressions between amplitude of sEMG and muscular force were linear. Figure 1 and 2 show this relationship for a typical subject V10 (female, age 22) with linear simple regression

equations as follows:

$$\text{Force} = 548.38 * \text{sEMG} + 0.66 \qquad \text{Flexor digitorum}$$

$$\text{Force} = 27.15 * \text{sEMG} + 0.39 \qquad \text{Extensor digitorum}$$

The regression coefficients R^2 are 0.957 and 0.640 for flexor and extensor digitorum accordingly. These regression coefficients for all 7 subjects are with means of 0.756 and 0.782 for flexor and extensor digitorum. These graphical and numerical representations identified a numbers of important results, which are outlined below:

(1). The sEMG increased for the ascending levels of contractile force for both muscles(flexor and extensor digitorum). This relationship may be well described as the linear with the relatively high regression coefficients(Table 1). There was no sudden change in sEMG signals during each trial, indicating consistent muscle contraction pattern.

(2). There are significantly larger slopes of regression lines for flexor than extensor digitorum (Table 1).

(3). Maximum amplitudes of muscle force are higher for flexor than extensor digitorum during maximum voluntary contractions(MVC) (Table 2).

The theory of muscle physiology states that muscle activity patterns are determined by the dominant forms of muscle fiber metabolism. It is widely known that the aerobic metabolism plays a major role in dynamic movements, while the anaerobic in isometric activity. In the former, the blood supply is abundant, while in the latter it is deficient.

It might be theoretically acceptable that a dynamic muscular activity can be composed of numerous isometric muscular activities within relatively short periods of time(i.e. 2 seconds). This short time insured the availability of aerobic metabolism for muscle cells. Therefore the results derived from an isometric study might be carefully lead to a related dynamic situation. Since the muscle force is not only a function of the sEMG, time and muscle fiber distribution but position as well, caution must be taken to pursue this kind of isometric-to-dynamic application. And the certain limitations must be imposed on this extrapolation with regard to the movement position and time duration. In general, a deliberate experiment design is necessary to accommodate a specified investigation. A study of finger force requirements of keyboard operators during typing will be presented separately in another paper.

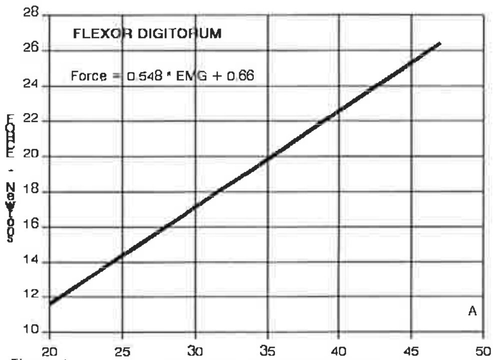

Figure 1

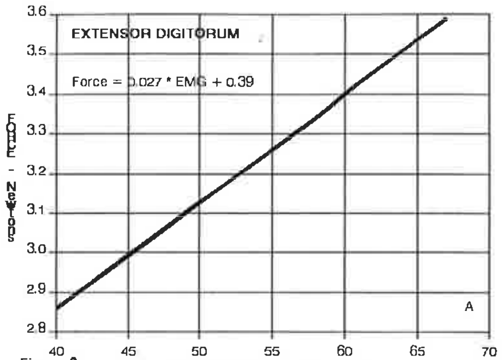

Figure 2

Table 1. The Regression Coefficients(R^2) and the Slopes(b) of Regression Lines:
Mean Values Plus/Minus Standard Deviation

	R^2	b*
Flexor Digitorum	0.756±0.145	239.9±146.8
Extensor Digitorum	0.782±0.159	53.2±51.3

* P=0.0354 (paired t-test).

Table 2. The Maximum Amplitudes of sEMG and Muscle Force:
Mean Values Plus/Minus Standard Deviation

	sEMG(mv)	Force(N)
Flexor Digitorum	0.275±0.160	20.91±11.75
Extensor Digitorum	0.361±0.285	7.77±5.43

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

3-D MOTION OF THE PELVIS DURING PASSIVE LEG LIFTING

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INTRODUCTION

The passive straight leg raising test (SLR) is frequently used to assess low back dysfunction and hamstring length. The limits to the range of motion during this test have been attributed to nerve-root pain, muscle spasm, muscle length, or mechanical dysfunction (1,2). Bohannon (3) in a two-dimensional cinematographic investigation of SLR measured the leg moving in the sagittal plane at a 2.7:1 ratio with the pelvis. Fisk (2) observed posterior movement of the anterior superior iliac spine (ASIS) which he associated with lumbar flexion and contralateral hip extension. Grieve (1) observed pelvic rotation in the sagittal plane to be coupled with a pelvic rotation in the transverse plane toward the non-lifted leg. No investigation has measured three-dimensional pelvic motions during the straight leg raising test to describe coupled motions in the pelvis . The purpose of this investigation was to describe the relationship between leg lift motion and pelvic motion in three-dimensions.

MATERIALS AND METHODS

Leg and pelvic motions were measured in three dimensions during passive SLR on 23 male subjects. Each subject was asymptomatic for low back pain and without significant low back or lower extremity dysfunction as determined by a screening exam (4).

A sonic digitizer (Model GP-8-3D, Science Accessories Corp.) was used to measure the three-dimensional positions of the pelvis and leg during right and left SLR. Pelvic motion was tracked via spark gaps mounted on a rigid aluminum plate which was positioned on the subject's pubic symphysis and bilateral ASIS's. Leg motion was tracked via three spark gaps each that were mounted on right and left aluminum frames. Each frame was strapped securely to the straight leg over an orthopedic knee immobilizer which maintained full knee extension during the leg raising test.

Each leg lift trial began with a two-second resting period. The leg was raised in approximately seven seconds and then lowered at the same rate. This rate, approximately 15°/sec, provided sonic data at approximately every 1.0° of leg motion.

The leg was lifted in the sagittal plane through its full physiological range of motion, or until the subject indicated the position caused pain. The leg was held in neutral hip rotation and neutral hip abduction/adduction. Three left and three right leg trials were randomized to minimize systematic errors.

The leg and pelvis rotation angles were calculated from the dot product between vectors representing the initial and final positions. The total leg lift motion was calculated in the sagittal plane, while angles of rotation for pelvic motion were calculated in the sagittal, transverse, and frontal planes.

RESULTS

Univariate repeated measures F-tests on the three right and the three left leg trials showed no significant differences in leg and pelvic motions. As a result, the values for the right trials were averaged and the values for the left trials were averaged. The average ranges of motion for the leg in the sagittal plane and the pelvis in the sagittal, transverse and frontal planes are presented in Table I.

TABLE I

LEG LIFT, PELVIS AND HIP ROTATION ANGLES.

Variable	Leg	Ave	SD
Max Leg Lift Angle	Rt.	59.3°	9.1
	Lt.	60.3°	8.9
Pelvic Angle: Sag. Plane	Rt.	16.9°	3.0
	Lt.	17.1°	3.3
Pelvic Angle: Tran. Plane	Rt.	1.6°	2.5*
	Lt.	-1.5°	2.6
Pelvic Angle: Front. Plane	Rt.	6.6°	1.8*
	Lt.	-6.5°	2.6
Hip Angle	Rt.	41.1°	8.5
	Lt.	41.2°	8.3

* $p \leq 0.005$

The maximum leg lift angle is the sum of hip and pelvis motions in the sagittal plane. Hip rotation in Table I, was calculated by subtracting pelvic rotation in the sagittal plane from maximum leg lift rotation. Paired sample t-tests on right versus left maximal leg lift angle, hip rotation angle and pelvic rotation in the sagittal plane showed no significant differences ($\alpha \leq .05$) between the two sides. Significant differences were noted between right and left leg lifts for pelvic rotation in the transverse and frontal planes.

The ratios of leg motion in the sagittal plane to pelvic motion in each of the three cardinal anatomical planes are described in Table II. The 3.5:1 ratio of leg to pelvic motion in the sagittal plane was consistent between right and left legs. However, the ratios of leg to pelvic motion in the transverse and frontal planes were significantly different as evaluated by a paired t-test.

TABLE II

RATIOS OF LEG LIFT TO PELVIC MOTION

Variable	Leg	Ave	SD
Leg Lift/Pelvic Sag.	Rt.	3.52	0.75
	Lt.	3.49	0.67
Leg Lift/Pelvic Trans.	Rt.	38.10	24.45*
	Lt.	-28.19	14.60
Leg Lift/Pelvic Frontal	Rt.	8.91	2.21*
	Lt.	-10.01	2.89*

* $p \leq 0.001$

DISCUSSION

Angular rotations for leg, hip and pelvic motion in the sagittal plane were lower than those reported previously (3,5). In the current study, knee extension was very rigidly maintained which probably accounts for the lower mean leg lift angle. However, the ratio of total leg lift angle of rotation to pelvic rotation in the sagittal plane was larger than expected. As previously indicated, Bohannon (3) reported a 2.7:1 ratio which may be explained by differences of measurement technique. Since motion in the transverse and frontal planes had not been measured, there are no data for comparison. In our opinion, however, the magnitude of pelvic motion in these other two planes is not as important as the direction.

Rotation of the pelvis in the sagittal plane occurred in the same direction as the leg being lifted while rotation of the pelvis in the transverse plane occurred toward the non-lifted leg. Rotation of the pelvis in the frontal plane was seen as a superior movement of right ASIS during the right leg trials and a superior movement of left ASIS during the left leg trials. The statistically significant differences in pelvic rotation in the transverse and frontal planes are due to differences in direction rather than magnitude of rotation.

In conclusion, motion of the leg during straight leg raising produces coupled motions of the pelvis that are similar to those observed during trunk sidebending and walking. Thus, the directions of rotations in the transverse and frontal planes during straight leg raising suggests that the motion of the pelvis applies a mechanical strain on tissues that are associated with low back pain. With further investigation, the measurement of three-dimensional pelvic motion may lead to more understanding and a better clinical test for individuals with low back dysfunction.

ACKNOWLEDGEMENTS

Funding for this project was provided in part by Johnson Controls, Inc. and a Biomedical Research Support Grant from the College of Osteopathic Medicine, Michigan State University.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

CONTRALATERAL COACTIVATION IN THE SHOULDER-NECK REGION

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INTRODUCTION

Contralateral coactivation (Coco) indicates, that voluntary activity in muscles of one body side induces unintended (co-)activation of muscles of the other body side. For decades Coco has been advanced as the theoretical background for methods rehabilitating paretic muscles, although previous attempts to show Coco in body limbs have been inconsistent (1, 2). Recently Schüldt & Harms-Ringdahl (3) demonstrated Coco of the shoulder muscles during unilateral maximal isometric contractions.

However, to our knowledge investigations have never been carried out concerning a) the quantitative relationship between Coco and ipsilateral load, and b) the dependence of Coco on body position. From an ergonomic point of view these issues are important especially as regards the muscles in the shoulder-neck region, a common site of load-related disorders. In addition, Coco is important in relation to basic functional anatomy and motor control.

The present study aims at assessing quantitatively contralateral coactivation of the upper trapezius (UT) muscle during ipsi- and bilateral glenohumeral (GH) torques with the arm(s) in different positions in front of the body. The study was part of a more comprehensive investigation of EMG activity in the shoulder-neck region according to arm position and GH torque (4).

MATERIALS AND METHODS

12 healthy volunteers (6m, 6f, age 24-45 ys) carried out isometric ramp contractions against gravity using their stretched arm (right, left, or both) in 8 arm positions in front of the body. EMG was obtained from one pair of surface electrodes placed above the descending part of UT. The simultaneous recordings of pull force and EMG were processed to give the regressed relationship between GH torque and EMG activity, the latter normalized (normEMG) with respect to a standardized 15 Nm test contraction. Finally, normEMG values corresponding to GH torques of 15 Nm and 30 Nm were calculated from the regression equation.

All ramp contractions were repeated on another test day, with a reversed order of arm positions.

Coco was quantified through R_{uni} and R_{bil}, defined in fig. 1.

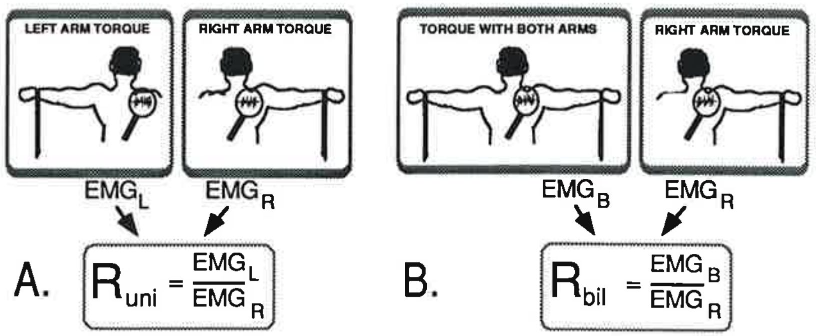


Fig.1. Definition of (A) the one-arm-only coactivation ratio R_{uni}, and (B) the bilateral coactivation ratio R_{bil}.

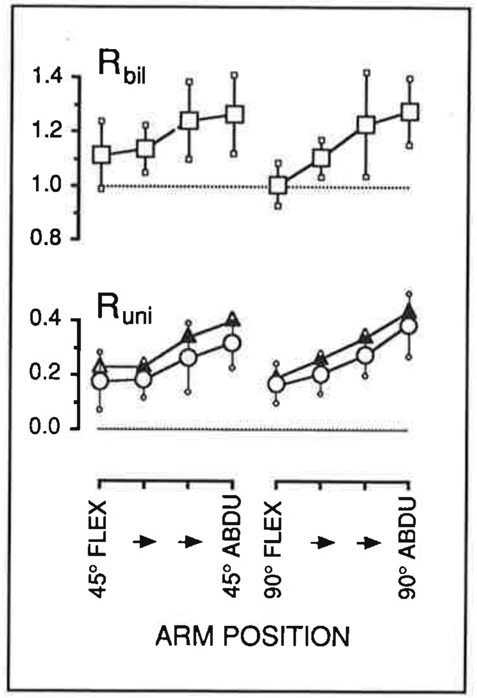
RESULTS

Values for R_{uni} and R_{bil} are presented in fig.2. A GH torque of 15 Nm with one arm only results in Coco ($R_{uni}>0$) in all arm positions (as shown by the CI's in fig. 2). Increasing the GH torque to 30 Nm increases R_{uni} significantly in most positions as shown in fig. 2. The size of R_{uni} relates to arm position ($p<0.01$, ANOVA), abducted positions giving more Coco than flexed. GH torques with both arms result in greater normEMG than unilateral torques alone ($R_{bil}>1$) in 6 out of 8 arm positions (c.f. the CI's in fig. 2), but this coactivation is not significantly related to the GH torque level (ANOVA). Thus, the R_{bil}'s presented in fig. 2 are means of the values obtained at GH torques 15 Nm and 30 Nm.

Adding a GH torque with the left arm to a right arm torque results in an increase in normEMG, which is less than the contralateral EMG activity induced by the left arm alone ($R_{bil}-1<R_{uni}$ in all 8 arm positions, $p<0.01$, t-test). Thus, Coco is not simply additive.

The mean values of fig. 2 conceal a substantial intra- and interindividual variation in Coco pattern. The within-subjects-between-days variation coefficient in R_{uni} was 70%. Torque-EMG relationships (ramps) differed in shape between subjects. This is illustrated in fig. 3. Neither R_{uni} nor R_{bil} was affected systematically by the number of trials performed, i.e. learning did not interfere with Coco.

Fig. 2. R_{uni} (O: 15 Nm; △,▲: 30 Nm) and R_{bil} (□) in relation to arm position. Mean values with 99% confidence intervals. ▲: value larger than the O-value in the same arm position (paired t-test, $p<0.05$).



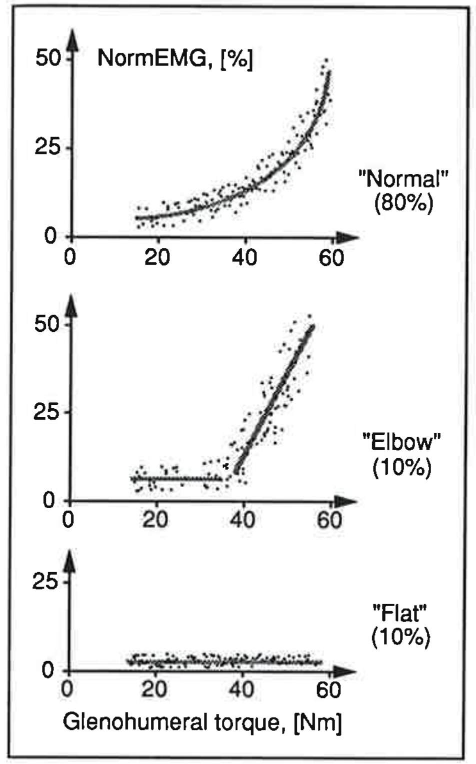


Fig. 3. Contralateral EMG in the right UT during a left-arm GH torque. Observed ramp types, and frequency of ramp occurrence. Total number of ramps recorded: 768.

DISCUSSION

Ergonomics

The Coco induced in the right UT muscle by a modest left arm torque may reach levels above the occupational guidelines for static muscle load suggested by Jonsson (5). Thus, analysis of working operations must take into consideration possible activation of "contralateral" body regions not voluntarily used by the worker, especially if the work comprises handling of heavy loads. Coco is suggested as a contributing etiological factor for the development of musculoskeletal disorders in upper body regions not primarily engaged in working operations.

Functional anatomy

Biomechanical models have been presented, estimating the activity of individual muscles in the shoulder region during manual work (e.g. 6). These models rely upon muscle activation being related only to the position and load of the ipsilateral arm. According to these models, EMG in the upper trapezius muscle is not affected by contralateral GH torques. Thus, our results dispute the validity of such biomechanical models.

The contralateral UT activity during a one-arm torque is sufficient to raise the shoulder against gravity. As, however, no shoulder movement is seen, the UT activation must be part of a cocontraction pattern around the contralateral shoulder.

Motor control

UT may be regarded mainly a postural muscle, elevating and rotating scapula during arm activity. The engagement of the contralateral UT during a one-arm GH torque may thus be viewed as part of a postural stabilization pattern, aiming at compensating instability in the frontal plane. This suggestion is consistent with the observation, that Coco is most prominent in abduction. The large inter-individual variation in Coco could reflect the plasticity in postural control of the shoulder region.

However, Coco is recorded also during the posturally stable bilateral GH torque. This may support the idea of Coco being a result of excitation overflow: the inappropriate involvement of irrelevant muscles, which is especially evident during forceful voluntary contractions. Excitation overflow may be strongly influenced by the subjects's motivation and concentration, and may thus explain the large observed between-days-within-subject variation in Coco.

ACKNOWLEDGEMENTS

This project was supported by the Swedish Work Environmental Fund (project # 89-0044).

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

INFLUENCE OF SELECTIVE TRAINING USING MYOFEEDBACK ON THE ELECTROMYOGRAPHY OF THE MM. VASTI MEDIALIS AND LATERALIS.

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INTRODUCTION

The use of electromyography (EMG) biofeedback as a means for treating conservatively patellofemoral pain syndrome and subluxation of the patella is well known in rehabilitation. It is based upon the hypothesis that selectively training the action of the M. vastus medialis (thus improving its strength) may result in a realignment of the patella into the femoral groove (Williams & Street, 1976; LeVeau & Rogers, 1980; Wild et al., 1982; Wise et al.,1984; Suzuki et al., 1985). However, the theory of selective action of the M. vastus medialis during extension of the knee is still controversial (Möller et al., 1986).

The purpose of this study was to evaluate electromyographic changes in the relation between M. vastus medialis (VM) and M. vastus lateralis (VL) after a "selective" training of the M. vastus medialis using EMG myofeedback.

MATERIAL AND METHODS

Ten healthy subjects (mean age: 26.8 (±5.7) years, body weight: 69.4 (±11.5) kg, body height: 174.6 (±7.6) cm) participated in a 3 week (5 days a week, 30 min) training program in which isometric actions of the VM were emphasized over the VL. Four sessions of 10 contractions, each contraction lasting 10 s, followed by a 10 s rest period, were monitored on BFS 50 (Medico Electronics, Belgium) using passive bipolar electrodes. The subjects were asked to lighten as many LED's (coming from the electrodes of the VM) as possible, and to keep the LED's from the VL as low as possible. Between each session, a rest period of 1 min was allowed.

Before and after the training program, the subjects were tested on an isokinetic dynamometer (Kin.Com, Chattex Corp., TN, USA) in the same position as during the training, i.e. knee extension in a sitting position.

Electromyography of both the VM and the VL, using the Electromyography Signal Processing and Analysis System (Cabri, 1989), was recorded by means of active bipolar surface electrodes (gain:10; CMRR 90dB) and additional amplifier (gain:100; CMRR:85dB), and stored on a FM recorder (Teac HR 30) for later analysis.

The pre and post training tests consisted of the following: (1) determination of the maximal isometric maximum (MVC), (2) three isometric contractions at a level of 30% and (3) 70 % of maximum (6s), and (4) three maximal isometric contractions (3s).

The electromyographic analysis included integrated EMG (iEMG, expressed as a percentage of the highest individual peak) and maximal amplitude (μV).

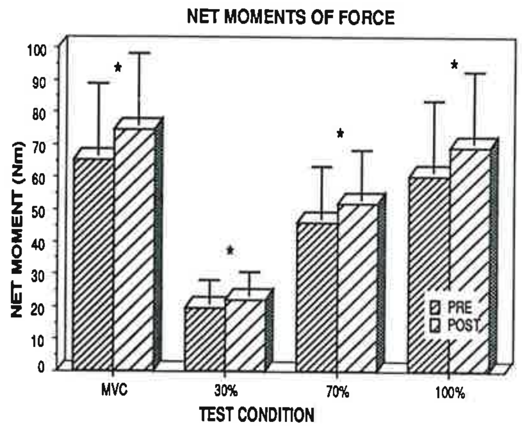

Figure 1: Mean net moments of force (±SD) at maximal voluntary contraction (MVC), 30, 70 and 100% of MVC. Asterisks denote significant differences (P<0.05).

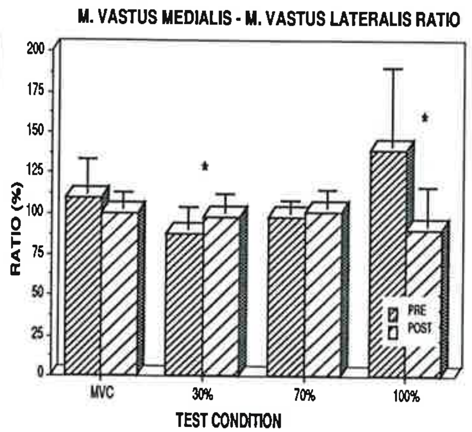

Figure 2: Mean VM/VL ratio (±SD) (calculated on the basis of the iEMG) at the different tests. Asterisks denote significant differences (P<0.05).

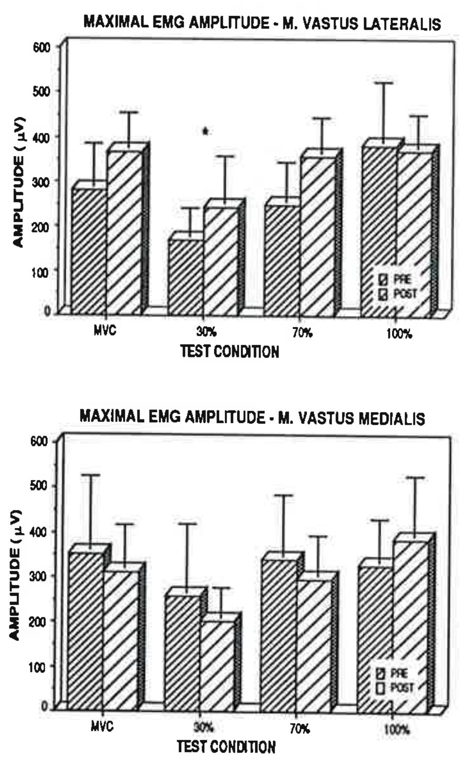

Figure 3: Maximal amplitude (mean values ±SD) of VL (top) and VM (bottom) at the different tests. Asterisks denote significant difference at P<0.05.

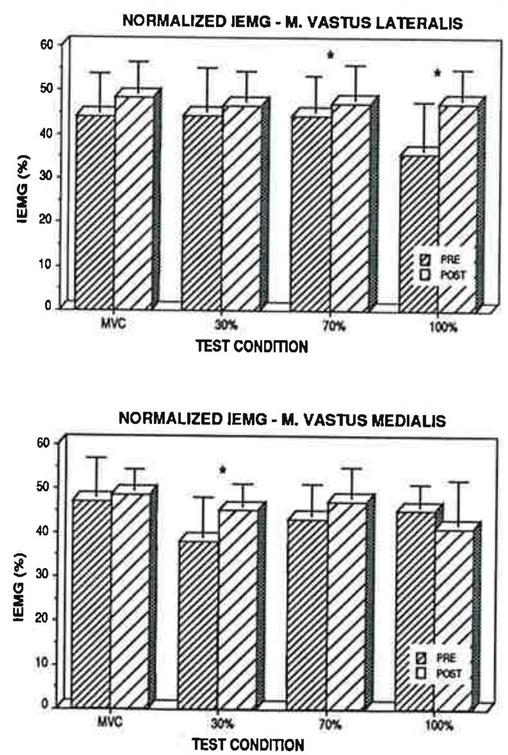

Figure 4: Normalized iEMG (mean values ±SD)of VL (top) and VM (bottom) at the different tests. Asterisks denote significant difference at P<0.05.

The statistical analysis consisted of Kolmogorov-Smirnov and paired Student-t statistics at a significance level of P<0.05.

RESULTS

The results (Table 1) of the *maximal net moment* measurements showed a significant increase (P<0.05) after the training period (Fig. 1), which was partially reflected by a significant increase of both iEMG and maximal amplitude of the Mm. vasti (Fig. 3 & 4).

TABLE 1
MEAN VALUES FOR NET MOMENTS, IEMG AND MAXIMAL AMPLITUDES (max amp) of the M. vastus medialis (VM) and M. vastus lateralis (VL), before (PRE) and after (POST) the training period at maximal voluntary contraction (MVC), 30, 70 and 100% of MVC. Numbers in italics denote ± 1 standard deviation.

	Net Moment (Nm)		iEMG VM (%)		iEMG VL (%)		MAX AMP VM (µV)		MAX AMP VL (µV)	
	PRE	POST	PRE	POST	PRE	POST	PRE	POST	PRE	POST
MVC	65.4	74.4	47.0	48.6	43.9	48.4	350.4	309.4	278.6	366.5
	22.2	*21.6*	*9.0*	*5.3*	*9.2*	*6.7*	*165.2*	*97.5*	*104.3*	*79.0*
30%	19.5	22.2	38.4	44.7	44.1	46.5	256,9	201.8	168.6	242.0
	6.6	*6.6*	*9.6*	*5.4*	*9.5*	*6.8*	*152.7*	*64.5*	*66.8*	*107.1*
70%	45.9	51.9	42.9	47.0	44.3	47.0	338.2	295.7	245.5	354.8
	15.6	*15.0*	*6.8*	*7.4*	*7.8*	*7.7*	*138.3*	*88.6*	*96.3*	*80.8*
100%	60.0	69.0	45.2	41.2	35.6	47.1	325,7	385.3	381.6	368,7
	22.0	*21.6*	*4.9*	*10.1*	*11.1*	*6.6*	*95,9*	*135.4*	*150.7*	*72.9*

Maximal EMG amplitude was significantly increased after the training period in the VM at 30% of maximal voluntary contraction (MVC). In all other cases, no significant differences were found (P>0.05).

When the *normalized iEMG* was considered, significantly higher values were found between pre- and post training sessions in the VL at 70% and 100% of MVC, and in the VM at 30% of MVC. The iEMG of the VM was only significantly higher from the VL before the training period at 100% of maximal effort (P<0.05). After the training period, the iEMG of the VL in comparison to the VM was lower at 30% and 100% MVC (P<0.05).

The ratio between iEMG values (Fig. 2) of VM and VL (*VM/VL ratio*; expressed as a percentage), only showed a significant higher ratio after the training period at 30% of maximal effort, and was significant lower at 100% MVC (P<0.05).

DISCUSSION

Correct alignment of the patella depends considerably on the muscle balance between the

VM and the VL (LeVeau & Rogers, 1980). With knowledge that the only selective function of the VM is patellar alignment (Basmajian and Deluca, 1985), it is thought that strengthening the VM independently from other parts of the M. quadriceps femoris may contribute to solve this problem.

Although others have reported that the use of EMG biofeedback training is an efficient and successful treatment approach for patients with patellofemoral pain syndrome (LeVeau & Rogers, 1980; Wise et al, 1984), our results did not demonstrate a conscious control of VM over the VL at all exercises. The VM only showed a significant increase at 30% MVC, whereas higher intensities did not show any significant increase of activity. This phenomenon is probably due to the active neuromotor control of muscle action at lower intensities. However, it may be questioned that exercising at low intensities will result in an effective training influence, and in this case it is pertinent to ask whether training at 30% of MVC only, will induce a realignment of the patella.

Maximal isometric moments are significantly increased, indicating a strengthening influence of the biofeedback training period. This is also reflected by the increase of iEMG of the VL at 100% of MVC and a lower value of VM/VL ratio, which is due to the increase of activity of that same muscle. Thus, it can be stated that the present training procedure has a strengthening effect, but that this influence is more pronounced in the VL than in the VM.

These data suggest that, under the present circumstances, "selective" training of the VM, thus separating its activation, by means of myofeedback cannot be regarded as a valid method in order to obtain an effective realignment of the patella. Furthermore, it has been demonstrated that this "selectivity" in activity is only valid in lower intensities of effort, in which case it may be questioned whether a remaining training effect exists.

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ERGONOMICS

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

ELECTROMYOGRAPHY IN ERGONOMICS

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INTRODUCTION

In ergonomics much interest has been focused on the load on individual muscles or muscle groups. One reason for this interest is the commonly accepted hypothesis that muscular overloading may result in inflammatory reactions in tendons or in their surrondings, or that muscular overloading may result in muscular fatigue which in turn will result in myalgic reactions. The former part of the hypothesis has to a great degree focused the interest on quantitative and qualitative electromyographic aspects of muscular load during work, while the latter part of the hypothesis has focused the interest on electromyographic muscle fatigue evaluation.

There exists a widely accepted ergonomic paradigm saying that high muscular load and muscular fatigue are injurious. On the other hand we know from sports physiology that high muscular load resulting in muscular fatigue is a prerequisit for increasing the muscular capacity. As a result there also exists a widely accepted sport paradigm saying that high muscular load and muscular fatigue are desirable in order to obtain a proper training effect. Basic muscle physiological research may help us to understand that both paradigms may be correct.

There are at least four important and common applications of electromyography within ergonomics:

- Muscle coordination studies
- Quantitative electromyography
- Muscle fatigue studies
- Qualitative electromyography

MUSCLE COORDINATION STUDIES

Muscular coordination patterns during different postures and movements may be mapped out by using a fairly simple recording equipment without facilities for quantifying the myoelectric signals. A common biofeedback equipment may be sufficient and adequate for this type of study. It should be pointed out, however, that in those cases where a simple biofeedback equipment is used for the EMG investigation the aim should be to restrict the questions to the problem of which muscles are involved.

A great number of ergonomic studies have dealt with the recording of myoelectric signals from different muscles in various work postures and work movements. In many of these studies the recording of postures or movements have been very short, often only a few seconds. Unfortunately the results from such short-term studies do not always reveal the effects of real work situations. The coordination pattern during well controlled laboratory tests may be different from that in a similar authentic work situation. The worker will change his posture during work and coordination patterns recorded in well controlled short-term laboratory studies may not be used in the real work situations. Stress during the work situation may cause muscle tenderness which does not occur in the laboratory study. As a consequence ergonomic muscle coordination studies have to be performed over long periods of time and in real work situations.

Recent studies have indicated that even minor interindividual differences in the muscle coordination pattern may be important factors in the development of or in the prevention from musculoskeletal disorders among workers. Thus muscle coordinaion studies may play an important role also in future ergonomic studies provided that these coordination studies are extended over long periods of time and that they are performed in real work situations.

QUANTITATIVE ELECTROMYOGRAPHY

Within ergonomic electromyography much interest has been focused on estimating the absolute load or the relative load on individual muscles during different work tasks. The background is the old and probably not quite correct ergonomic rule of thumb that any decrease in the muscular load during work may be an advantage.

There is always a background noice in the myoelectric recordings. This noice may interfere with the exact estimation of very low levels of muscular force. This means that electromyography may be of limited value for the exact quantitative estimation of those very low static load levels which are common in contemporary work situations.

A very common way to normalize the myoelectric signal amplitudes is to express the EMG amplitudes in per cent of the mean amplitude recorded during a maximal voluntary static contraction. The quantitative estimation of the relative force of contraction during those moderately high forces of muscular contraction which are common in many work situations may then be very unprecise in those cases where there is not a linear relationship between the myoelectric signal amplitude and the relative force of muscular contraction.

Electromyography has a limited value also for exact quantitative estimation of the absolute or relative load levels in strong contractions. Very often the peak loads will be estimated to be more than 100 per cent of maximal voluntary contraction.

In ergonomics there may not always be a need for exact measures of the relative or absolute muscular force of contraction. The ergonomist may on the other hand very often need to know if the muscular load is higher in one work situation compared to another work situation. In these cases quantitative electromyographic evaluations without any attempts to normalize the signals may be fully sufficient.

MUSCLE FATIGUE STUDIES

Localized muscle fatigue is accompanied by characteristic changes of the myoelectric signal parameters such as an increase in the myoelectric signal amplitude and a decrease in the mean power frequency.

It is well known that an increase in the myoelectric signal amplitude may be either an effect of increased muscular performance or of muscular fatigue. It has also been shown that the mean power frequency will decrease with decreasing force of contraction for those low levels of contraction which do normally occur in many work situations.

As a consequence there is no single electromyographic parameter which can be used as the only indicator of the degree of local muscular fatigue. It has been shown that it is often easy to follow the myoelectric fatigue reactions in well controlled laboratory experiments but many scientists who have tried to use similar methods in applied ergonomic studies report technical difficulties in obtaining reliable results when trying to identify signs of muscular fatigue by analysing the myoelectric signal parameters of the varying or intermittent myoelectric signals recorded during the work task.

QUALITATIVE ELECTROMYOGRAPHY

Long-lasting static loading, in particular that of neck and shoulder muscles, poses a significant ergonomic problem in the management of contemporary occupational situations. The static load level may indeed be the most important factor causing muscular fatigue during constrained work, much more important than the mean load level. For that reason the evaluation of the static load level may be an important factor in the analysis of myoelectric signals recorded during a period of work.

The lowest level of muscular activation recorded during a work period may be referred to as the "static" component of muscular activation.

From the amplitude distribution curve it is among other things possible to calculate the approximate level of the "static" component of muscular activation. The static load level may then for practical reasons be defined as the lowest amplitude present during at least 90 per cent of the total recording time.

CONCLUDING REMARKS

The future role of electromyography in ergonomics will be closely linked to the future new roles and concepts of ergonomics.

Changes in the work organization rather than technical improvements of the work places will probably be the main future task for the ergonomists. Job rotation, job enlargement and changes in the work schedules will be the main means for the ergonomists. Great attention will be paid to the duration and the distribution of pauses during work. The main task will be qualitative changes in the work load. Increasing the work load may not always be considered to be a disadvantage in future ergonomics. As a consequence qualitative electromyography examined over extended periods of time will probably be the most important area of ergonomic electromyography, much more important than quantitative electromyography and muscle fatigue studies.

Even minor interindividual differences in the muscular coordination patterns during work may be important for the development of musculoskeletal disorders. For that reason it may be important in future ergonomic electromyography to perform detailed coordination studies in different work situations and to pay special attention to the interindividual differences.

Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

ERGONOMIC EVALUATION OF HAND LEVERS FOR MACHINE OPERATORS

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INTRODUCTION

During the last thirty years, there has been a continuous mechanization of the Swedish forestry industry. The operation of today's forestry machines requires intense work with hand levers in repetitive short-cycle movement patterns and fixed sitting positions with the arms and hands engaged during 50 up to 90 - 95 % of the working time (2). This is considered to cause tendinitis or tendovaginitis in the shoulders.

In order to decrease the muscle load in the necks and shoulders of the operators, possible causes for the disorders when operating levers have been identified in ergonomic investigations both in the laboratory and in the field. The results from the laboratory studies have been field tested and later introduced to the manufacturers for improvements and been used in developing guidelines (3). The most important of these studies concern optimum work posture, optimum working range for hands and arms, lever resistance, lever movement spans, work intensity and the effect of an arm rest, fixed as well as movable. The purpose of the investigations has been to reach results useful not only in forestry machines, but also in other types of similer equipment.

Methods for simulating the operation of hand lever controls in the laboratory have been developed. Muscle load was studied by means of electromyographic methods. The amplitude distribution of the rms of the EMG-signal proved to be useful for the estimation and comparison of muscle activity in different work situations and postures. In most studies, efficiency and subjective ratings have been registered and calculated together with the muscle load analyses. Below, five laboratory investigations where EMG-methods have been involved are summarized.

METHODS AND RESULTS

Optimum work range and muscle load in the operation of hand levers. In an investigation of optimum work range in operating levers, subjects determined the best location of the levers for three different work heights (fig1). In order to survey the muscle load in the shoulder when operating a lever, nine shoulder/arm muscles were then examined by electromyography on subjects, performing in these positions and the three different work heights. The results disclosed that the field of the optimum range usually took the shape of an ellipse. Both the EMG and the subjective choices resulted in relatively small ranges for comfortable work (fig2). Lever locations providing an elbow angle of about 105° when the upper arm was parallel to the trunk proved to be the most comfortable position studied. However there were individual variations which were not ascribable to differences in body size (1).

Fig 1 Determination of optimum work range. A pen is fixed to the lever to mark the optimum position.

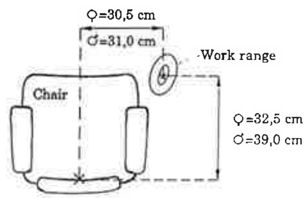

Fig 2 Small work range of comfort gave the ellipse a major axis of 70 mm. A large range gave the ellipse a major axis of 160 mm

Muscular load with different degrees of hand control resistance

Muscular load in the shoulder was studied with the subjects in a sitting position operating the hand lever. The simulated work was designed as a computor aided tracking task on a monitor (fig3). EMG-activities in the trapezius and deltoid muscles were recorded, and movement precision was measured as the time taken to move a cursor to a number of targets on the video screen. Three different degrees of control resistance in the ranges 5 - 10 N, 6 - 12 N and 7 - 14 N respectively were used. With the strongest lever resistance, a higher muscular load than with low or medium resistance was observed in the posterior and middle parts of the deltoid muscle, than for the other two. The choice of resistance seemed to be less important for the load on the trapezius muscle and the anterior part of the deltoid muscle in this study. The precision in operation did not seem to be as good with the lowest lever resistance as with the other two, though the difference was not statistically significant. A biomechanical model was presented in order to estimate the level of muscular force about the shoulder joint. The largest muscular force calculated, is roughly

equivalent to 30 - 40 % of the force necessary to hold the arm in a horisontal position (5).

Different ranges of deflection in hand lever operating.

Two different ranges of lever motion, 0-35 mm and 0-60 mm, were studied in simulated work in a computor aided tracking test with a hand lever control as described above (fig3). The results showed a significantly lower EMG-activity in the trapezius muscle when the smaller range of lever motion was used. The change of motion range had no statistically significant effect on the EMG-activity of the other muscles studied, deltoid, flexor and extensor carpi radialis muscles. The investigation showed that movement precision was remarkably better (P=0,0001) when the smaller motion range was used (6).

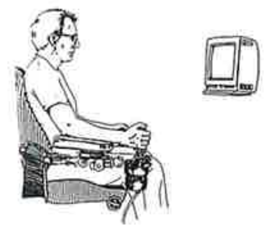

Fig 3 Test arrangement in studies of resistance and deflection.

Effect of elbow rest during hand lever operation.

In a previous study it was observed that the use of an arm support in forward- backward movements might increase the load on the shoulder musculature. This incited an investigation of the possible load relieving effect of an elbow rest on neck and shoulder muscles during lever operation. The subjects were studied during the operation of a lever control in a simulator with and without an arm rest. The motion of the arm was examined by means of video recordings, and muscle loads were estimated by collecting the EMG-signal from the upper trapezius and the middle and posterior parts of the deltoid muscle. The video recordings revealed different movement patterns for different subjects. To what extent the load on the neck and shoulder muscles can be reduced will depend on how well the operator manages to adapt his movements to the elbow rest and its position. No significant difference in EMG-activity was found, but a decrease of the activity was recorded in the trapezius except when the lever was pulled backwards. The activity in the lateral and posterior part of the deltoid muscle increased when the elbow rest was used, particulary when the lever was pulled backwards (4).

EMG evaluations of fixed and movable arm rests when operating hand controlled levers, in simulated tests.

In a pilot investigation not quite concluded, the muscle activity was registered in nine shoulder and arm muscles during work with support from two different arm rests. An arm rest following the forward - backward movements of the arm was compared to an ordinary fixed one when working with a simulated test controlled by a hand lever. Preliminary results show a tendency towards a lower EMG-signal in the trapezius muscle but a somewhat higher activity in the lower arm muscle flexor radialis. Six out of eight subjects preferred the movable arm rest because of the feeling of lower load in the shoulder and ease in handling the control. No difference in precision and efficiency was registered (Unpublished).

In conclusion our laboratory and field investigations together with several follow up studies, show that improvements of equipment design, work environment and work organization will decrease the muscle load in forest machine operators. The improvements should be concentrated to the areas of:

* excessive work intensity during work in fixed, ergonomically inappropriate sitting positions
* repetitive, short cycle movement patterns
* controls offering excessive resistance
* controls with excessive movement spans

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

STATIC WORK LOADS AND OCCUPATIONAL MYALGIA-A NEW EXPLANATION MODEL

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INTRODUCTION

Static muscle loads in occupational work have been considered inappropriate since long. There is today strong epidemiological evidence that static workloads are related to muscular disorders, at least in the shoulder region [1], [6], . Considerable effort has been spent in the industry to prevent these disorders by reducing the static load levels, according to the traditional view (see below). Little success has been accomplished by this approach.

TRADITIONAL MODEL

The traditional model for the etiology of these muscular disorders is based on findings and models emerging from laboratory studies of endurance times at different static load levels [9]. The intramuscular pressure increases with increasing muscle contraction which impedes a sufficient circulation. This deficient circulation is believed to cause muscle disorders [8], [2]. This approach implies that it is possible to find a static load level which is low enough to allow unlimited contraction duration without risking muscle disorders.

CONTRADICTIONS OF THE TRADITIONAL MODEL

One reasonable consequence of the traditional model is that individuals with low maximal force capacity would be at higher risk for muscle disorders than stronger individuals doing the same job. Several investigations have failed to show such a relation [5], [4], [12].

Another reasonable presumption is that the degree of electromyographic signs of fatigue during work would predict these disorders. Hägg et al failed to show any such relations in a longitudinal study [4]. Furthermore, muscular disorders have been reported at occupational static load levels as low as 1%MVC [11]. At such a low load level it is not reasonable to expect insufficient circulation.

ALTERNATIVE MODEL

The proposed model is based on two corner-stones. The first one is the fixed recruitment order of motor units discovered by Henneman and colleagues [3]. This principle is described schematically in figure 1. When motor units are recruited to

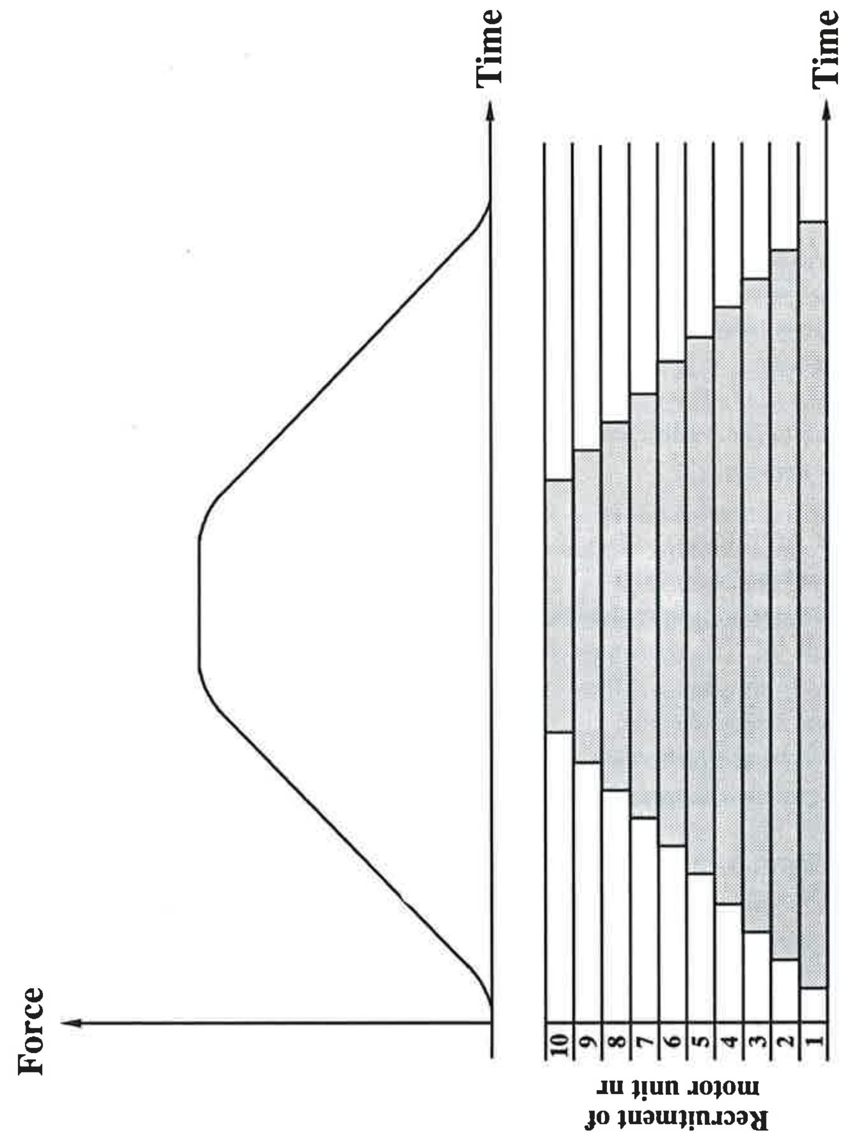

Fig. 1. Schematic demonstration of the ordered recruitment according to Henneman. Top curve: Force developement. Bottom curve: Recruitment.

build up the muscle force, this is done according to a fixed order which implies that low threshold motor units (type I) always are recruited first and remain active until total relaxation of the muscle.

The second corner-stone is findings by K G Henriksson and colleagues when studying muscle biopsies from patients with muscular disorders related to occupational static loads. They found that certain type I muscle fibres seem to be injured *selectively* ('ragged red fibres') [7]. The nature of these fibre injuries is interesting but this question is left out here as it does not concern the model as such.

Considering these two facts, it is reasonable to believe that the fibres belonging to the bottom motor unit of the pyramid ('Cinderella') in figure 1 is affected first, as a result of too long activation and too little rest. If the load conditions are not changed, motor units higher up in the pyramid are affected successively. This process develops slowly over several months or maybe years.

DISCUSSION

The alternative model explains why the maximum force capacity plays no role for the development of these disorders. A stronger person certainly utilizes a smaller fraction of the total muscle capacity compared to a weaker person doing the same job. However, according to the Henneman size principle, the low threshold motor units of the two persons are activated to the same degree.

Traditional electromyographic fatigue studies becomes irrelevant for the prevention of these disorders due to the fact that surface EMG represents the compound contribution from a large number of motor units. However, when the number of affected motor units gets large enough as in a manifest myalgic disorder, increased electromyographic signs of fatigue are seen during work [10] [4].

This proposed model implies important consequences for ergonomics. Reduction of static load levels is not effective as a preventive measure as far as these disorders are concerned. Instead the duration of the load and the pauses should be more considered. However, our knowledge is still too limited to be able to suggest any recommendations.

Even if the traditional model is irrelevant for the understanding of the disorders discussed here, it is quite valid when discussing endurance times and related topics at higher static load levels.

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ELECTROMYOGRAPHICAL STUDIES ON CHECK-OUT WORK

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INTRODUCTION

Automatic price recording units which employ scanners to read the bar codes on goods are increasingly being used at check-outs. Price recording via scanners is very different from the method at conventional cash registers where data are input using keyboards. Since the pattern of hand/arm movements at both types of cash desks varies, a difference in the strain of the musculature involved can be expected. Electromyographical measurements were carried out in laboratory and field studies in order to compare the muscle strain at different cash desks and to determine possible muscle fatigue.

METHOD

The field study involved 6 female cashiers at scanner check-outs in a supermarket. They were each examined on a day-long basis during a six-day working week. In the laboratory study 6 female cashiers performed a 60-minute simulation of cash-desk work at both a conventional cash desk and a scanner check-out. Fig. 1 provides a schematic representation of the scanner check-out used in the field study in the supermarket. In the laboratory investigation two cash desks of a type similar to that in Fig. 1 were employed, one with a scanner, the other without.

In all of the investigations the electromyographical signals from 4 muscles in the left shoulder (m. deltoideus, pars acromialis and m. trapezius, pars transversa) and the left arm (m. biceps brachii and m. extensor carpi ulnaris) were derived using bi-polar surface electrodes and recorded on magnetic tape. The muscles in the left half of the body were selected because the majority of goods were transferred using the left hand. In addition, surveys reveal that a high incidence of complaints occurs in the left shoulder/arm region for work at both conventional cash desks [1, 2] and scanner check-outs [3].

An opto-electrical method was additionally employed to record the movements of the left wrist. The person's activity was recorded

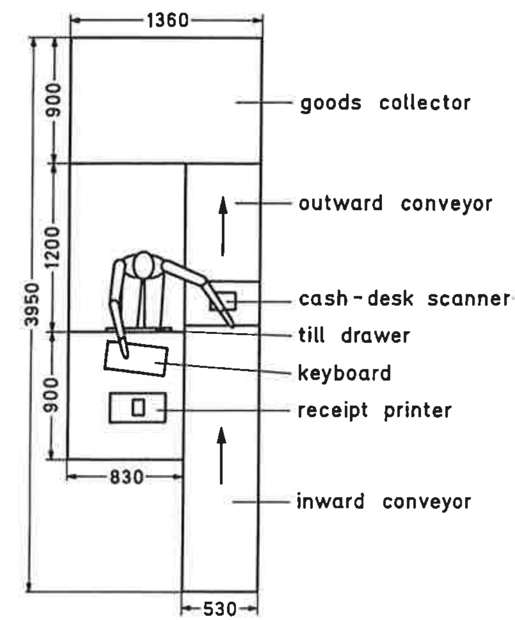

Fig. 1: Schematic representation of the cash desk in the field study; dimensions in mm

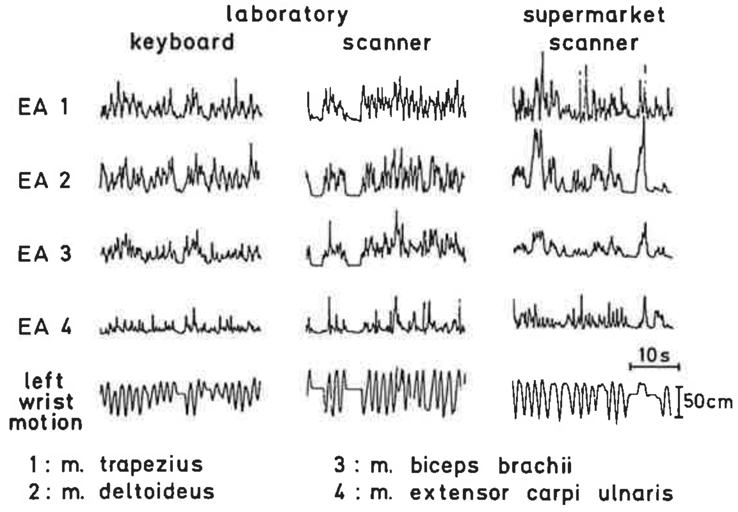

Fig. 2: Original traces of the electrical activity (EA) and the left wrist motion from the laboratory and field studies

by means of an "electronic protocol" as described by [4]. The protocol was produced by an observer who used a numerical keyboard to input the actual activity of the cashier according to a predetermined code. The action code was recorded parallel to the physiological signals on the same tape. In the evaluation the Electrical Activity (EA) was determined from the electromyographical signals by means of rectification and analogous integration (time window of 200 ms).

RESULTS

Fig. 2 shows three short sections of recordings of the Electrical Activity and the left wrist motion from the laboratory and field studies. Comparison of the recordings reveals that the wrist, and thereby the goods too, are moved over a longer distance at the scanner check-outs in the laboratory (middle recording) and in the supermarket (right-hand recording) than in the case of price recording via a keyboard (left-hand recording). The transfer distances are determined by the dimensions of the cash desks and scanners. An evaluation of the EMG signals from the laboratory study demonstrates that the myoelectrical activity is higher for most of the muscles at the scanner check-out than at the conventional cash desk. The increased muscle activity is mainly due to the longer transfer distances for goods at the scanner check-out and to the need to lift each article over the scanner.

A field-study result is shown in Fig. 3. The mean EA during scanning and the number of scanned articles per min are indicated for a whole working day. Each dot in the upper diagrams corresponds to the mean EA in a section in which goods are scanned. The scanning periods are determined using the action code. Regression analyses reveal that a significant increase in EA occurs for at least one muscle in most of the working periods. Since there is no significant change over time in the number

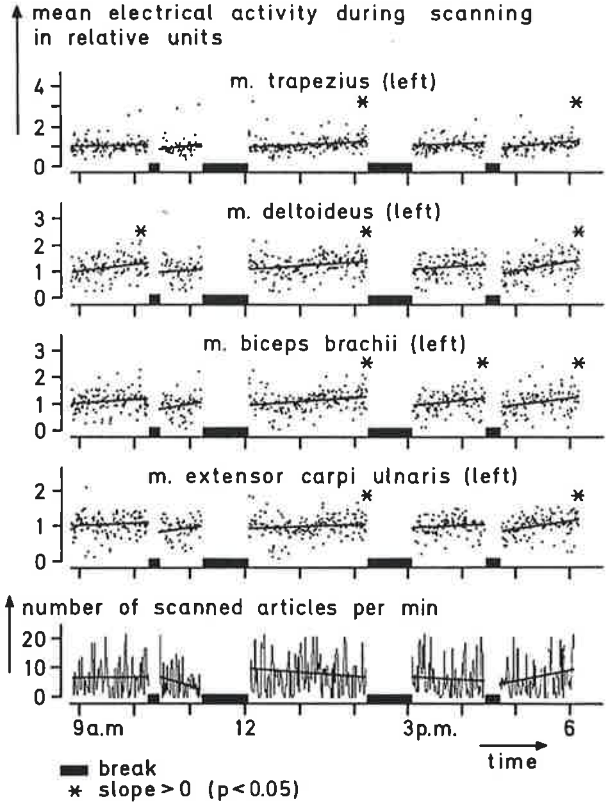

Fig. 3: Mean EA during scanning and the number of scanned articles during a whole working day in a supermarket

of goods per minute, it can be assumed that the workload is constant and that the increase in EA results from muscle fatigue. The increases in EA for all of the muscles examined during all the working periods from Tuesday to Saturday in the week of the investigation are collated in Fig. 4. Closed symbols are utilized when the slope for EA over time is significantly different to zero ($p < 0.05$). Positive fatigue-related increases in EA as a function of time are found above all on days during which the number of scanned goods per minute and the number of heavy goods per day are higher than average (e.g., Wednesday and Saturday).

RECOMMENDATIONS FOR WORK DESIGN

The design and the installation of the scanner in the cash desk should exclude long transfer distances for the goods during scanning. This can be achieved either by mounting the scanner vertically or by using narrow scanners so that the inward and outward conveyors (see Fig. 1) can be brought as closely together as possible, this reducing to a minimum the manual transfer distance for the goods.

Whenever muscle fatigue occurs, work can only be performed for a limited period and rest breaks must be provided. The connection between the slope of the fatigue-related increase in EA and the endurance time for fatiguing muscular work is described quantitatively by [5]. Accordingly, the possible duration of uninterrupted work can be predicted and the timing of a necessary break determined, on the basis of the EA increase. The right-hand ordinate in Fig. 4 shows the endurance time according to [5] for the slope of the EA provided on the left-hand ordinate. A minimum endurance time of approx. 50 min is found for the EA slopes measured in the field study on check-outs. On the basis of the findings, disregarding the last section on Saturday, limiting uninterrupted cash-desk work to between 90 and 100 min is recommended for purposes of practical application [6].

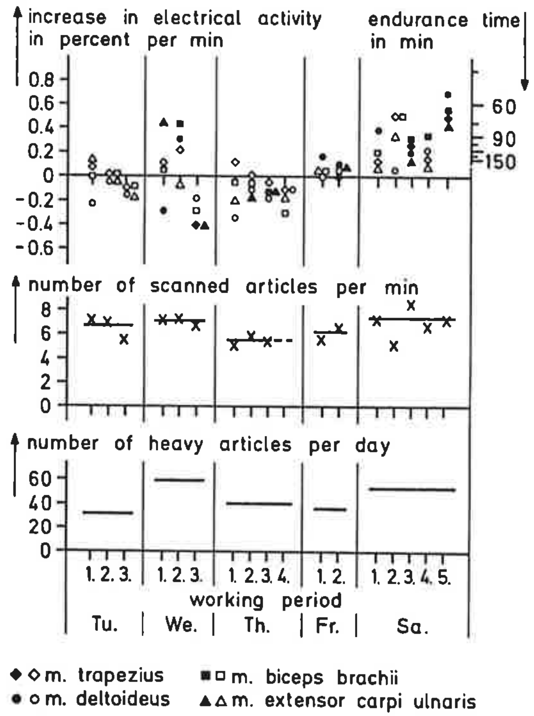

Fig. 4: Slope of EA time courses for all working periods from Tuesday to Saturday (upper diagram), number of scanned articles per minute (middle diagram), and the number of heavy articles per day (lower diagram)
closed symbols: slope of EA $\neq 0$; significant: $p < 0.05$;
open symbols: non-significant

CONCLUSIONS

The present study on check-outs demonstrates that electromyography is a suitable method for determining muscular strain and fatigue both in the laboratory and under shop-floor conditions during whole shifts. Recommendations for the design of scanner check-outs and for a physiologically justifiable system of breaks are derived from the electromyographical findings.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

MOVEMENT AND CHOICE-REACTION TIME TESTS AS INDICATORS OF OCCUPATIONAL MUSCLE LOAD

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INTRODUCTION

A basic assumption in ergonomic litterature is that workers performing monotonous or repetitive work tasks at a high level of muscle load are more at risk of developing musculoskeletal complaints than when the load level is low. However, subjects performing identical work tasks may show considerable variation in muscle load (1). Thus, many workers with high load levels should be able to perform their work at a much reduced muscle load, without changing the physical layout of their work space. If such 'high-risk' subjects could be identified, an appropriate training program may be successful in reducing muscle tension at work.

The present study attempts to relate the variation in occupational muscle tension to two recognized sources of variation in muscle usage: motor efficiency when performing dynamic movements and stress-induced tension components appearing at minimal body movement. A machine-paced work task was selected for the study, to eliminate differences in work pace as a source of variation in muscle load.

MATERIAL AND METHODS

Thirtynine female workers, age 21 to 65 years and with at least 2 years experience in their present job, were included in the material. The work task consisted of removing chocolate bars from a chocolate coating machine, or ready-packed chocolate from chocolate packing machines. The work tasks were machine-paced and demanded continuous movement of the arms with the working area situated just below elbow height. There was no indication of differences in static or median load level (2) depending on the work task, and the material was therefore not stratified with regard to this parameter.

Shoulder muscle tension was assessed by surface EMG recordings from the descending (upper) trapezius muscles during work. The EMG signals were full-wave rectified and calibrated in per cent of the EMG activity at maximal voluntary contraction (% MEMG). The EMG equipment and experimental set-up were the same as in Westgaard[3].

EMG activity was also recorded during a test with continuous movement of the dominant hand between three target areas, and during a complex reaction time test which was performed with minimal and symmetric body movement. The purpose of the two tests, which are more fully described by Westgaard and Bjørklund[4], was to obtain some indication of the level of muscle activation during dynamic movement with a demand of muscle coordination, as well as the tendency of the subject to react with muscle tension to tasks which require a high level of attention, but no body movement.

The statistical analysis was performed as linear regression analyses, using the SPSS statistical package.

RESULTS

Median trapezius muscle load during work showed considerable variation, between 0.6 and 12.1 % MEMG for the shoulder with the highest load level. A similar variation in median muscle load was observed both for the movement test (1.1 to 17.5 % MEMG for the active trapezius, 0.1 to 9.2 % MEMG for the passive trapezius) and

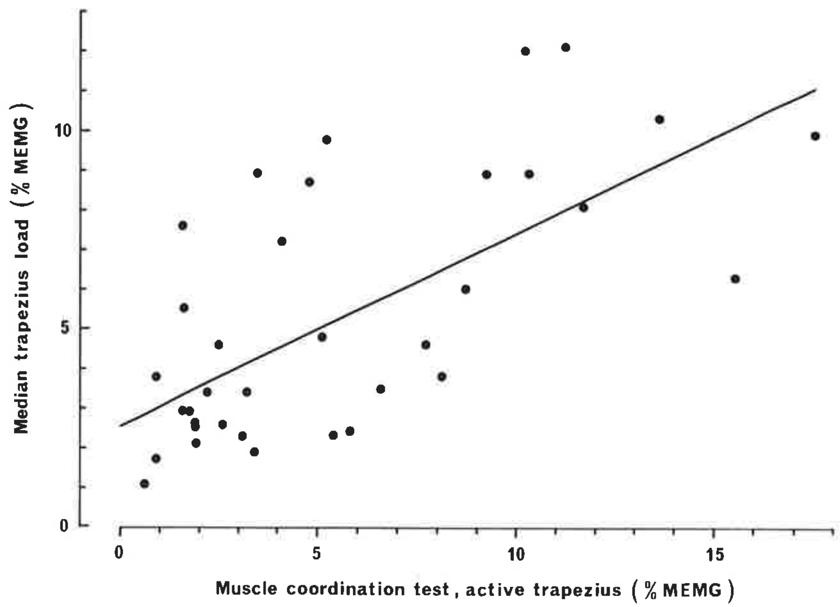

Fig. 1. Median trapezius load at work as a function of median load of the active trapezius during the movement test.

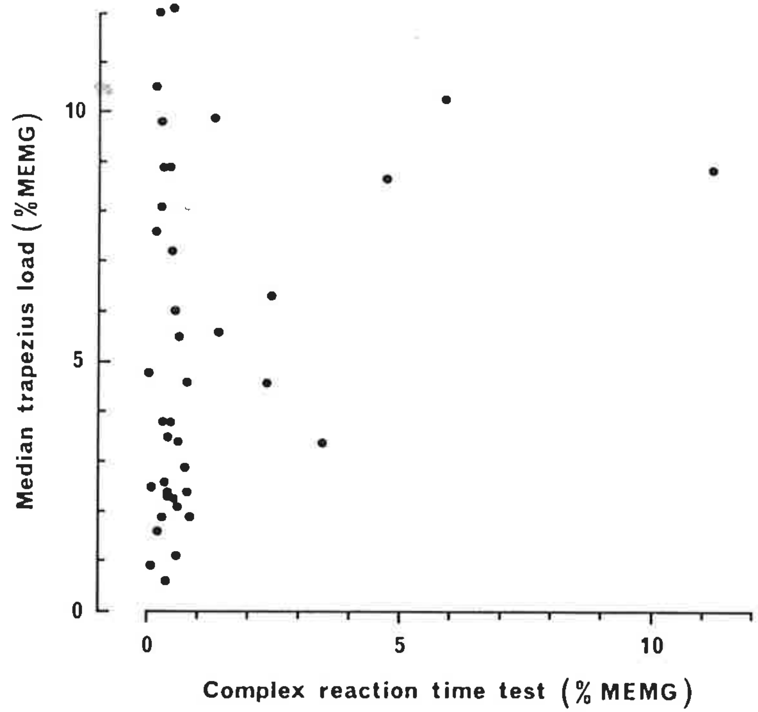

Fig. 2. Median trapezius load at work as a function of median trapezius load (average of both muscles) during the complex reaction time test.

for the complex choice reaction test (between 0.1 and 11.2 % MEMG, as a mean of the left and right trapezius).

A strong correlation between median median muscle load of the active trapezius during the movement test and median muscle load at work was found ($r=0.65$, $p<0.0001$; Fig. 1). There was also a significant, but less strong correlation between median muscle load at work and the passive trapezius during the movement test ($r=0.47$, $p<0.002$).

In contrast, there was no correlation between muscle load at work and tension during the complex reaction time test. However, as seen in Fig. 2, those subjects who responded with a relatively high level of muscle tension, >2 % MEMG, also recorded high load levels at work. Subjects with low trapezius loads in this test recorded any level of trapezius load at work. Thus, high load levels during the complex reaction time test may be a possible indicator of high vocational load, but the reverse conclusion is not valid.

When comparing the movement and complex reaction time test, it was found that trapezius tension during the choice reaction test correlated strongly with tension in the passive trapezius during the movement test (r=0.76, p<0.0001), but not with tension in the active trapezius.

DISCUSSION

Very little is known about the activation of the musculoskeletal system in normal, everyday activities. However, it is generally recognized that motor skills vary widely. Muscle tension is also an integral part of intrinsic arousal mechanisms, whether this is a consequence of anxiety, performance demands, interpersonal relationships or other stressors in the environment (5). Thus, activation mechanisms not readily observed may well contribute to the sometimes considerable prevalence of musculoskeletal complaints in the general population and in occupations without any postural demands.

In the present study we attempt to quantify these sources of muscle tension by the application of two relatively simple tests. Much work remains before the relationship between the tests and the underlying muscle activation mechanisms is understood. However, it is encouraging that clear correlations are emerging between some of the parameters used in the tests and occupational muscle load within a very limited experimental material. Thus, it appears that the movement test correlates well with occupational muscle load for a work situation demanding fast, repetitive movements. Also, responders at the complex reaction time test appear at risk of developing a high level of occupational muscle load. Similar studies of other occupational groups will now be performed, to examine the generality of these conclusions.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

SYSTEMATIC DIFFERENCES BETWEEN THE ELECTROMYOGRAPHIC FATIGUE INDICES MEAN POWER FREQUENCY, MEDIAN FREQUENCY AND ZERO CROSSING RATE DURING OCCUPATIONAL WORK

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INTRODUCTION

The shift of EMG spectra towards lower frequencies at fatigue, here refered to as electromyographic signs of fatigue (ESF) is well known [6]. To quantify this shift, several spectrum shift estimators have been proposed like median frequency (MF) [9], mean power frequency (MPF) [6] or zero crossing rate (ZC) [3]. A complete understanding of the spectrum alteration is still lacking but it is evident that a decrease of motor unit action potential velocity (MUAPV) is an important factor [6]. If a general MUAPV decrease would be the only effect, the *relative* decrease of the three estimators mentioned above would be the same. However, Broman et al [1] and Sadoyama et al [8] reported a far from ideal correlation between these three estimators and MUAPV which indicates other influencing factors as well. None of these authors analysed their data to reveal possible systematic differences. This work is carried out to further elucidate these circumstances under occupational working conditions. The null hypothesis is that there are no systematic differences between these estimators of ESF.

DATA COLLECTION

Previously collected EMG recordings during occupational work were reanalysed [10]. EMG was recorded percutaneously from the trapezius and infraspinatus muscles bilaterally during 10 s long test contractions (TC) [4]. A test contraction gives a load in the range 11-18% MVC. A test contraction was performed during short intermissions (approx. 25 s) of ordinary assembly work every tenth minute over a 2 hour period of work. All subjects were female. They all worked sitting and they performed either a coil winding task or an assembly task demanding high concentration. See [10] for further details. 39 muscle recordings from 14 subjects were analysed.

DATA ANALYSIS

The first 6 s of all TC recordings were analysed with a Brüel & Kjær 2032 FFT spectrum analyser and simultaneously the zero crossing rate was determined according to [3]. The spectra were transfered to a PC where MF, MPF and also the expected number of zero crossings, E[ZC], were calculated according to Rice [7]. (See below under Discussion for analytic expression.). All these readings were normalized to the reading of the initial TC for each muscle.

RESULTS

All absolute readings were correlated yielding the correlation matrix in table 1. The normalized readings of each fatigue index were averaged over the 39 muscles. These results are shown in figure 1. It should be noted that the rank order of the three indices is the same at all points. Given the null hypothesis, the probability of obtaining such a result is

$$p = \frac{1}{(3!)^{(12-1)}} < 10^{-8}$$

Hence there is strong reason for rejecting the null hypothesis.

Table 1.Mutual correlation coefficients for all absolute readings

	MF	E[ZC]	ZC
MPF	0.91	0.93	0.90
MF		0.73	0.70
E[ZC]			0.95

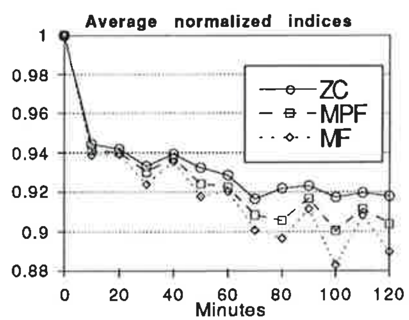


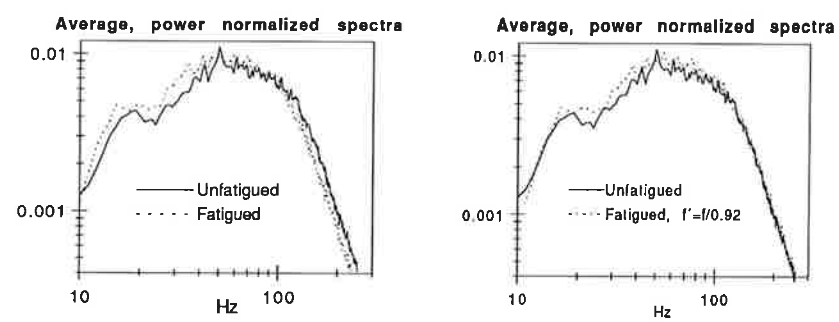
Fig. 1. Average values of the normalized indices of 39 muscle registrations

DISCUSSION

In order to search for a possible explanation of the systematic differences, all spectra from the initial TC were normalized to unity power and averaged. The same procedure was performed with all spectra from the final TC (after 2 hours of work). These 2 average spectra are shown in figure 2a. On the high frequency side, the difference between the two spectra could be well described as a true parallel shift which is in accordance with the simple MUAPV decrease model, proposed by Lindström [6]. On the low frequency side though, this simple approach seems to be less relevant. To estimate the average MUAPV decrease, the fatigued spectrum was shifted to the right by trial and error to an optimal fit in the high frequency region. A shift factor of 1/0.92 seems to be optimal. This modified plot is seen in figure 2b. The major remaining discrepancy lies in the range 20- 40 Hz. This fact can explain the systematic differences in figure 1. Regard the analytic expression for E[ZC] according to Rice [7]

$$E[ZC] = \left[\frac{\int_0^\infty f^2 \cdot S(f)\,df}{\int_0^\infty S(f)\,df}\right]^{1/2}$$

and compare them to the corresponding expressions for the other indices. (They are left out here due lack of space. See [9] and [6].) All three will show the same relative



a b

Fig. 2. Average spectra from the first (unfatigued) and the last (fatigued) TC. In b the fatigued spectrum has been shifted by a factor 1/0.92 to the right.

alteration due to the velocity decrease. The low frequency increase, however, will have the greatest relative impact on the MF, a little less on MPF and hardly at all on ZC due to the different powers of the frequency weighting of their analytic expressions. MF involves spectrum moments of order 0, MPF of order one and ZC of order two. The conclusion is that ZC alterations almost solely reflect the MUAP velocity decrease while MPF and, to a still greater extent MF, also respond to low frequency alterations related to firing statistics alterations (see [2]). It is also remarkable that the 'trial and error factor' 0.92 is very close to the final relative ZC value 0.917 (see figure 1). This observation supports the hypothesis that ZC reflects almost solely the MUAPV decrease.

The good correlation in table 1 between ZC and E[ZC] indicates that the hardware zero crossing counter readings follow the theoretically expected values, calculated from the spectra, very closely.

A more detailed presentation of these data will follow [5] where also increasing indices of ESF will be discussed.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

EFFECTS OF SIMULATED LIGHT REPETITIVE WORK

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INTRODUCTION

During the last decades the characteristics of work load in the industries have remarkably changed from heavy dynamic load to highly repetitive light work or light static loading of neck an shoulder muscles. Many contemporary work situations in the industries are characterized by repetitive movements, constrained postures, monotonous and boring tasks, regulated pace of work, lack of autonomy and of responsibility, distressed and uneasy feelings, etc. (1, 2). In spite of a seemingly good ergonomic work place layout and a remarkable reduction in the total work load, musculoskeletal disorders have become one of the most serious occupational problems in industrialized countries (3, 4, 5, 6).

Repetitive movements seem to be one of the most important hazardous factors in many types of industrial work (7, 8, 9).

The aim of the present study was to evaluate the effects of light repetitive work on muscular fatigue by means of electromyography, muscle tenderness threshold estimation and ratings of subjective discomfort in the musculoskeletal system.

MATERIAL AND METHODS

Material

Thirteen healthy young women, average age 24.9 years, volunteered in the study. Ten of them were professional cleaners. None of them worked professionally with repetitive industrial work.

Procedure

The subjects were asked to continuously perform a highly repetitive work for two hours without any pauses except for brief test contractions. The task was to insert small nails and small screws on plates (size 155 x 85 mm with holes for 120 nails and 45 screws, respectively).

In order to identify myoelectric signs of muscular fatigue standardized test contractions were carried out for 10 seconds every 20 minutes during the two hours work task.

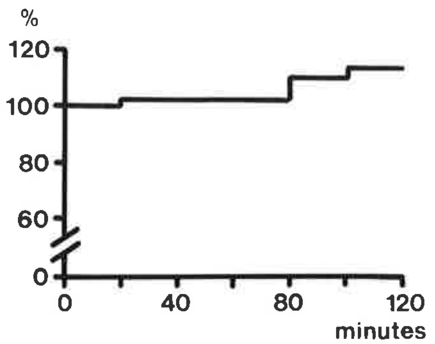

Fig. 1. Mean EMG amplitude of the right trapezius muscle during the work task. Values are expressed as a percentage of the value of the initial 20 minutes work task.

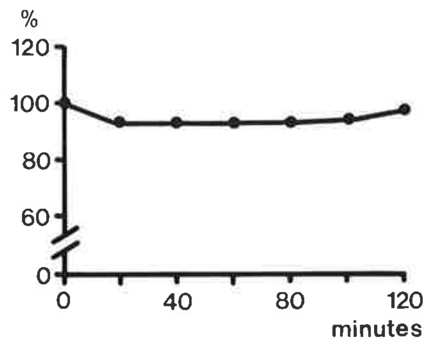

Fig. 2. MPF of the right trapezius muscle during the test contractions. Values are expressed as a percentage of the initial contraction value.

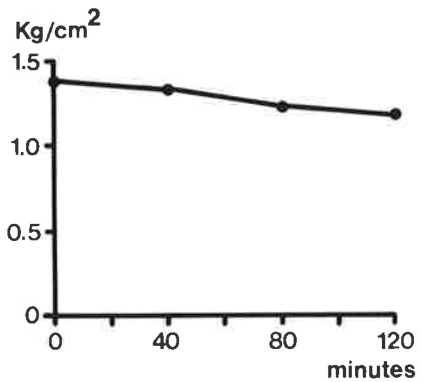

Fig. 3. Muscle tenderness threshold of the right trapezius muscle during the work task.

Electromyography

Electromyographic signals were recorded bilaterally from the descending part of the trapezius muscle, and from the infraspinatus and extensor carpi radialis longus muscles of the dominant side of the subject. Bipolar surface electrodes were placed in parallel with the muscle fibers. The myoelectric signals were transmitted by telemetry and recorded on a FM tape recorder.

The root mean square (RMS) values of the myoelectric signals during the work task and the mean power frequency (MPF) of the myoelectric signals during the test contractions were analysed using a computer.

Muscle tenderness threshold

During the work task the tenderness threshold of the trapezius muscles was measured by a small strain-gauge sensor. The tip of the sensor was pressed against the trapezius muscle with increasing pressure from 0 up to a maximum of 4 kg/cm^2 until the subject reported that she felt tenderness.

Rating of subjective discomfort

The subjects were also asked to rate their subjective discomfort such as stiffness, pain and numbness in different body regions during the work task using Borg's RPE scale.

RESULTS

During the two hours work task including the test contractions any clear increase in RMS (Fig. 1) or decrease in MPF (Fig. 2) indicating local muscular fatigue were not observed.

On the other hand the tenderness threshold level in the trapezius muscles (Fig. 3) decreased, and the subjective discomforts in the neck, shoulders and back increased during the two hours continuous repetitive work. Some of the subjective discomforts remained even until the next morning. The rates of subjective discomforts during the work task corresponded well with the tenderness threshold.

CONCLUSIONS

The results seem to indicate that continuous light repetitive work may be associated with subjective discomforts in the musculoskeletal system without being associated with electromyographic signs of muscular fatigue.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

TOWARDS A FIELD USABLE BIOMECHANICAL TYPOLOGY OF WORK

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INTRODUCTION

Arbouw has the task to improve the working conditions in the construction industry. In fulfilling this task it is essential to know where improvement is needed. Based on opinions of experts, sick leave percentages, surveys of occupations and other sources, it is decided to reduce the biomechanical workload in some risk occupations (see table 1). These risk occupations show mainly musculoskeletal problems, but also large cardiovascular and psychological problems. A project workload was developed to find the health hazards on these three fields. Part of this project workload, the method to find the major musculoskeletal workloads, is described in this paper. With this method field data are converted into a biomechanical typology of work activities.

TABLE 1

SICK LEAVE PERCENTAGE IN SOME BUILDING OCCUPATIONS IN THE NETHERLANDS IN 1988 (Source EIB, 1990).

occupation	n*	% sick leave	% musculo-° skeletal
gypsum bricklayer	600	14.3	55
scaffolding builder	2600	15,9	51
bricklayer's assistant	1400	12,3	55
steelbender	3000	13,9	53
digger	6000	12,3	50
total building industry	196000	10	49

*number of workers in the occupation in The Netherlands
°percentage of musculoskeletal problems of the total sick leave

BIOMECHANICAL TYPOLOGY

Selection of tasks. First, for each occupation the main tasks, the working methods, the used materials and tools are described. With a group of workers, employers, ex-workers and investigators the tasks, which consume most of the time are selected and light activities are excluded from further research. The steps towards improvement including data-capture, analysis and output are summarized in figure 1.

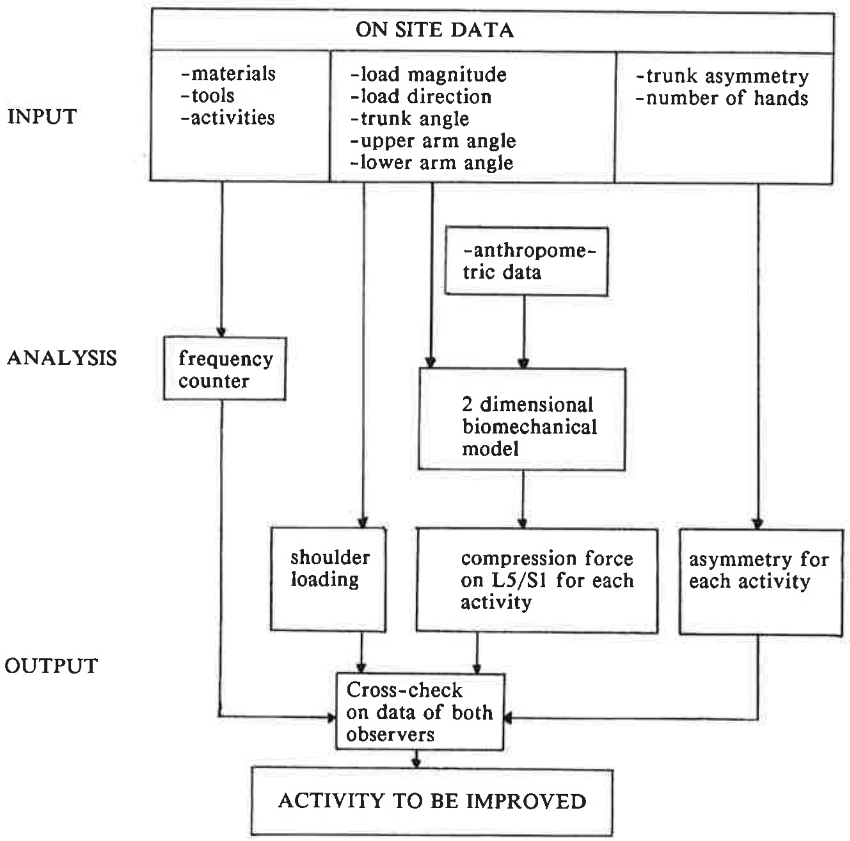

Fig. 1. Steps towards a biomechanical typology

Data capture. The selected tasks are recorded by the ROTA-system, which is a portable micro-computer for on-site data capture[4]. Elements of actions and postures are recorded 2500-3000 times by two observers in an occupation (6 subjects are recorded during 8 workactivities of approximately 20 minutes; 4 recordings/min). The keyboard of the micro-computer is divided into distinct sections. For each occupation a division of keys using keyboard overlays is defined before on-site data capture. To describe fully and accurately the work posture in biomechanical terms would take more time than 15 seconds. Therefore two different keyboard overlays are made to be used by two observers simultaneously. One observer assesses the trunk posture and the other assesses the position of the arms. Other elements are keyed in by both observers. These concern elements of the activity, gross body posture and weight of the object. Elements of the activity are recorded in combination with the biomechanical typology to find the tool, material, activity or movement which has the highest loading and needs to be improved. The weight of the object is also recorded by both observers, because this is difficult to estimate. Before analysing a specific profession the weights of the used materials and tools are defined on the keyboard overlay. However, observing remains difficult for instance in estimating the weight when parts of the gypsum bricks are used.

Analysis. A computer program has been developed that enables counting of frequencies and durations of tasks and further processing. Part of the data are led towards a two dimensional biomechanical model (2DSSPP model of the University of Michigan, version 4.0c) and for each activity, material, tool or posture the compression force on L5/S1 is calculated after the input of anthropometric data in the program. Based on a previous study[5] the trunk torsion, lateroflexion in combination with flexion is classified in risk postures.

A literature survey showed that two or more shoulderelevations larger then 60 degrees in a minute increase the risk of shoulder problems substantially [1,2,3]. The program searches for the observed shoulderelevations higher then 60 degrees. If these are found frequently during a specific activity, additional data can be put in the program to estimate the frequency more realistic.

Priority in improvement is given to those activities, which have the highest compression forces, which occur often, show most trunk asymmetry and shoulderload.

Errors. Two sources of errors are associated with the method:

1. errors due to recording consecutively.

The risk of making this error is reduced by making many recordings.

2. errors due to the level of detail and accuracy achievable in observing.

The program cross-checks the data of both observers. If a different score in the same class occurs, the computer displays the disagreement and gives the opportunity to change data. Furthermore the inter-observer agreement is calculated for those keys found in both overlays.

RESULTS

Inter-observer agreement. For the first three investigated occupations (gypsum bricklayer, scaffolding builder and steelbender) the inter-observer-agreement was always larger than 90%.

Compression force. For the gypsum bricklayer the compression force is in 11.1%, 1.0% and 0.5% of the observations higher than 2 kN, 2.5 kN and 3 kN respectively. The high compression forces were due to the trunk flexion (10.1% of the time a flexion of 75 degrees or more is observed) in combination with large weights of the gypsum bricks (180 - 210 N). A possible improvement is the setting out of gypsum bricks on a table (or car) to reduce the compression force during an activity which consumes much time.

Trunk asymmetry. In 3.6% of the observations asymmetries of 30 degrees or more were found.

Shoulder load. The shoulder is elevated 60 degrees or more in 15.4% of the time. Especially during bricklaying of the upper rows. This is also an activity which should be improved.

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REHABILITATION METHODS

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

REHABILITATION METHODS

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INTRODUCTION

In May 1976 the general assembly of the World Health Organisation agreed to the recommendation that a classification should be developed of impairments, disabilities and handicaps as an addition to the International Classification of Diseases, ICD. In 1980 the International Classification of Impairments, Disabilities and Handicaps was published by the World Health Organisation (1).

Disease may result in spontaneous or man-induced recovery or death. Recovery may be incomplete and leave residual impairments. Impairments may be reduced by the use of pharmacological or surgical methods. If still remaining impairments interfere with the desired level of functional ability, some form of disablement is present. The consequences of illness or trauma, or, rather, the disabilities they cause, are the starting point for physical and rehabilitation medicine (2).

Physical and Rehabilitation Medicine is a speciality which has the role of coordinating and ensuring the implementation of all the measures needed to prevent or minimise the functional, physical, psychical, social and economic consequences of impairment or disability.

It requires methodical steps to achieve these aims, from the time the deficiency appears to the time the patient can resume his/her place in both his/her own background and in society.

IMPAIRMENT, DISABILITY, HANDICAP

Illness-related phenomena are consequently:
disease —> impairment —> disability —> handicap.
These dimensions of the effects of disease need to be defined:

Impairment: (Déficience, Schädigung, Stoornis, Deficienza):
any loss or abnormality of psychological, physiological or anatomical structure or function.
The elementary functions of the locomotory, circulatory, respiratory, urogenital, digestive and neural system are impaired.
Disability (Incapacité, Beeinträchtigung, Beperking, Incapacità):
any restriction (resulting from an impairment) or lack of ability to perform an activity within the range considered normal for a human being.
The patients functional capacity is disabled. The functional limitations can be grouped around problems in areas of fields of attention:
A. Activities of daily living.
So. Social aptitudes.
P. Psychological aptitudes.
C. Communication aptitudes.
Handicap
A disadvantage for a given individual resulting from an impairment or a disability that limits or prevents the fulfilment of a role that is normal for that individual (depending on age, sex, and social and cultural factors) (3).
The handicap is judged in relation to evaluation of impairment and disability described above, in the context of the particular situation or restriction pertaining to the individual patient.

MEASUREMENT OF IMPAIRMENT AND DISABILITY

In modern rehabilitation a need is felt for methods for objective and quantitative evaluation of therapies and functional aids. Such methods can contribute to a more effective provision of rehabilitation care (4).
Inherent to their nature, objective and quantitative tools are most appropriate in the somatic field. The techniques currently available (generally developed in kinesiologic and biomechanical laboratories) can only provide information on the impairment level. Specifically in rehabilitation, a therapy or functional aid aims at the reduction of disability (the patient's functional capacity). Confining measurements to the impairment level means that the primary purpose of therapy (reduction of disability) is not studied.

In 1981 Brand and Crowninshield commented as follows on criteria for patient evaluation tools (5): diagnose distinguishes between diseases. Biomechanical tests are not helpful for distinguishing different diseases. They help to determine the severity of the disease or evaluate one parameter of the disease. In their opinion any patient evaluation tool will be useful only if the measured parameters correlate well with the patient's functional capacity. The clinical problem is that two patients with exactly the same diagnose and clinical symptoms can have a totally different functional capacity.

In the course of our attempts to apply gait analysis techniques in our clinical work, we became aware of these limitations of objective and quantitative techniques. Gait analysis has not become widespread outside the research institutions and useful for clinicians because it does not measure disability but only some parameters of the impairment. With increasing experience over the years, our views changed.

Pathology expresses itself not only in abnormal patterns or gaits, but also in absence of a part of such a range or of adaptability as such. Clinical gait appraisal should operationalize adaptability in a test of locomotor disability.

Such tests would pertain possibly not only to walking straight on over an even floor. Performances like stepping on and off a sidewalk, negotiating stairs, moving about while performing other tasks are possible items to enhance the validity of gait lab testing. Such tests would be directed to the disability-level in ICIDH terms or to the level of functionality: what a person is able to perform.

Disabilities will be measurable in future when researchers have developed new instruments of a more general character or even developed from social sciences and not from biomechanics (6).
Evaluation of disability on the basis of objective and quantitative measurements of impairments is generally not possible, because there is no unambiguous relation between impairment and disability.

As an example of our new approach, recently we started to develop tools to measure the functional aspects of a walking system prescribed to spinal cord injury patients using FES combined with a RGO (Reciprogating Gait Orthosis). We used the International

Classification of Impairments, Disabilities and Handicaps as a guide for the development of the evaluation instrument. Measurement of the functionality of walking systems means measurement of a number of related aspects. With respect to the measuring method they can be divided into two categories.
The first category consists of aspects that can be measured by means of objective instruments or semi-objective observation (comparable measurements have been performed before). Items in this category are the patient status, descriptions of the walking system and training method, and the patient's skills.

Aspects of the second category are subjective aspects that must be collected by means of patient inquiries. These subjective judgements are quantified using the Likert Scale technique (7).

The evaluation instrument described is still under development. In parallel we applied its principles in a pilot study to evaluate the activities performed with the RGO. The pilot study was part of a larger study, the goal of which was to direct the research on orthotic systems.

The model of Biomedical Technology Assessment is used. One of the conclusions is that the user of the technique (the patient) determines the degree of the success of the prescribed walking system.

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Electromyographical Kinesiology
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COMPARISON OF SPINAL MOBILITY AND STRENGTH TO EMG SPECTRAL PARAMETERS IN IDENTIFYING LOW BACK PAIN

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INTRODUCTION

Traditionally, joint motion and muscle strength have been used to identify individuals with low back pain (LBP).[1] Physical therapists have used spinal mobility as an objective clinical assessment of spinal function and back pain severity. Clinically, spinal mobility as described by forward bending (FB), backward bending (BB) and lateral bending (LB), has been employed to assess dysfunction and progress with rehabilitation.[2]

Recent advances in electromyography (EMG) indicate that there is an identifiable muscular component to chronic LBP. Individuals with LBP have been correctly identified solely on the basis of electromyographic spectral parameters.[3] These studies used the Back Analysis System (BAS) to assess changes in muscle fatigue characterized by EMG spectral changes. The median frequency (MF) of the power density spectrum has been shown to be a prime indicator of fatigue. The MF is defined as the frequency that divides the EMG power density spectrum into portions of equal energy. An integral part of the BAS is the Muscle Fatigue Monitor (MFM^{TM}) which tracks the MF of the EMG signal in real time during a sustained isometric contraction.

The purpose of this study was to compare the ability of traditional methods (spinal mobility and trunk extensor strength) to identify individuals with LBP to EMG spectral analysis in a combined athletic and non-athletic population.

METHODOLOGY

Subjects

Twenty-five members of the Boston University men's freshman crew team volunteered as subjects for this study. Descriptive profile of the crew members are presented in Table (1). All rowers were in the first month of training for the fall season. Rowers were classified as having LBP if they had a single or recurring incidence of lumbar LBP during the past year which interfered with activities of daily living, including rowing or training activities.

TABLE 1 : DESCRIPTIVE PROFILE OF CREW MEMBERS

CREW TEAM GROUP	AGE (yrs.)	HT (cm.)	WT (kg.)	MVC (kg.)	ROWING EXPERIENCE (yrs.)
Non-LBP	18.1	188.6	82.5	117.8	0.9
n = 17	(0.6)	(4.7)	(6.3)	(19.1)	(1.5)
LBP	18.8	189.3	82.4	114.8	2.4
n = 8	(1.0)	(2.7)	(7.5)	(24.2)	(3.3)

Values are mean with standard deviation in parentheses; HT = Height; WT = Weight; MVC = Maximal Voluntary Contraction

Procedures

Trunk range of motion (ROM) measurements were taken using inclinometers for FB and BB. Standard goniometric methods were followed for LB and rotation measurements. [4] Right and left LB were measured. A double arm full circle goniometer was used to measure trunk rotation (ROT).

EMG was detected via six active bipolar surface electrodes positioned bilaterally on the longissimus thoracis muscle at L1 spinal level, on the iliocostalis lumborum muscle at L2 spinal level, and on the multifidus muscle at L5 spinal level. The subject was positioned in the postural restraining device of the BAS. The following isometric contractions were produced: 1.) A three second maximum voluntary contraction (MVC); 2.) A thirty second contraction, at 80% MVC, was performed to induce fatigue in the low back extensor muscles; 3.) A ten second contraction, at 80% MVC, was performed at one minute into the recovery period following the fatiguing contraction.

Data Analysis

A two-way stepwise discriminant analysis was performed using ROM and MVC variables. This analysis determines how well these variables, when combined, discriminate rowers with LBP from those without LBP. For the EMG measurements, three parameters were calculated from the data for statistical analysis: 1.) SLOPE defined as the time rate of change of the MF; 2.) The initial median frequency (IMF) defined as the y-intercept of the linear regression described for SLOPE; 3.) The percent recovery (REC) of the MF at one minute.

A two-group stepwise discriminant analysis procedure was conducted separately for data from the fatigue trial and a single recovery trial to determine the optimal combination of MF parameters to classify LBP rowers from non-LBP rowers. The dependent variables included the IMF, SLOPE and REC parameters from the six electrode sites.

RESULTS AND DISCUSSION

The results of the discriminant analysis for the ROM and MVC variables are displayed in Figure (1). 63% of the non-LBP rowers were correctly identified. Of the total number of LBP rowers, discriminant analysis resulted in a correct classification of 57%. Right ROT was the only variable used for this analysis. This corresponded to three false negative classifications for non-LBP and six false positive classifications for LBP.

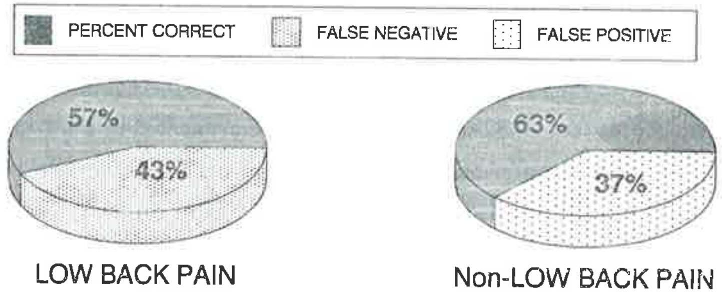

Figure 1: **Results from the ROM and MVC discriminant analysis.**

The results of the discriminant analysis for the MF parameters are displayed in Figure (2). The parameter that was the strongest discriminator was the percent recovery from the right L5 electrode. 100% of the non-LBP and 88% of the LBP rowers were correctly identified. This corresponded to one false negative classification for the non-LBP and no false positive classifications for the LBP.

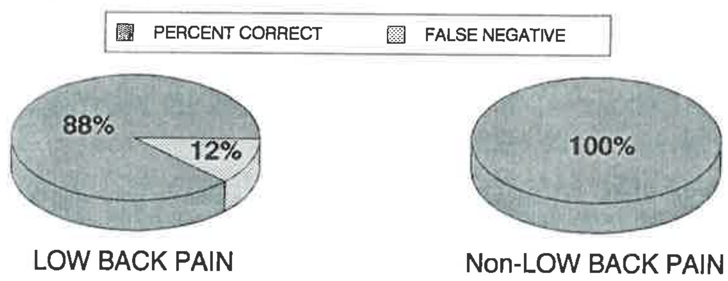

Figure 2: **Results from the Median Frequency discriminant analysis.**

These findings imply that while ROM and MVC can be used as indicators of progress with rehabilitation, they cannot be used effectively to identify an individual with LBP.

This study demonstrates that MF parameters can correctly identify rowers with and without LBP from a mixed population of non-athletic and athletic individuals. This finding verifies the results of Roy et al[5] who concluded that recovery at the L5 lumbar level was the best discriminating variable in identifying LBP among rowers.

CONCLUSION

Clinically, this study demonstrated that spinal mobility and isometric trunk extensor strength can be used to evaluate ROM and strength deficits and to measure rehabilitation progress. However, the ROM and strength variables were shown to be poor discriminators in identifying LBP.

This study confirms that EMG spectral analysis as implemented by the BAS can correctly identify LBP within a mixed population of untrained and trained freshman rowers. The right REC variable from the L5 level was the strongest discriminator used.

ACKNOWLEDGEMENT

This research was supported by the Rehabilitation Research and Development service of the Veterans Affairs.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

THE EFFECT OF SURGICAL PROCEDURES FOR CORRECTING ACL DEFICIENCY ON EMG PATTERNS DURING LOCOMOTION

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INTRODUCTION

Knee stability relies upon muscular action as well as ligamentous stability. The anterior cruciate ligament (ACL) is a major supporting structure in the tibial-femoral joint and surgical procedures are used to improve the mechanical stability of the joint when the ACL is ruptured through injury. The effect of two procedures, an extra-articular (EX) repair and a combined intra-articular and extra-articular (IN) repair, on the activity of the muscles acting around the knee joint have been investigated.

MEASUREMENT METHODOLOGY AND ANALYSIS

Twenty-six uninjured subjects, ten ACL deficient subjects with IN repair and eight subjects with EX repair were studied during level walking at free and fast speeds. Foot contact signals and electromyographic (EMG) signals of six major muscles have been measured and the linear envelopes (LE) of the EMGs in each muscle of each subject were calculated. The muscles studied were; rectus femoris (RF), vastus Lateralis (VL), vastus medialis (VM), semitendinosus (ST), biceps femoris (BF), and medial gastrocnemius (GS). The magnitude of each LE was normalized by its average over the stride. The stride was divided into 5% intervals. The distribution of the normalized magnitudes of the LEs at each of these intervals were compared between the normal and the surgical populations using a nonparametric statistical technique [1].

RESULTS

Comparisons of the populations of LE patterns show significant differences between normal and surgically repaired groups in all of the muscles. The changes incurred by ACL injury are also listed so

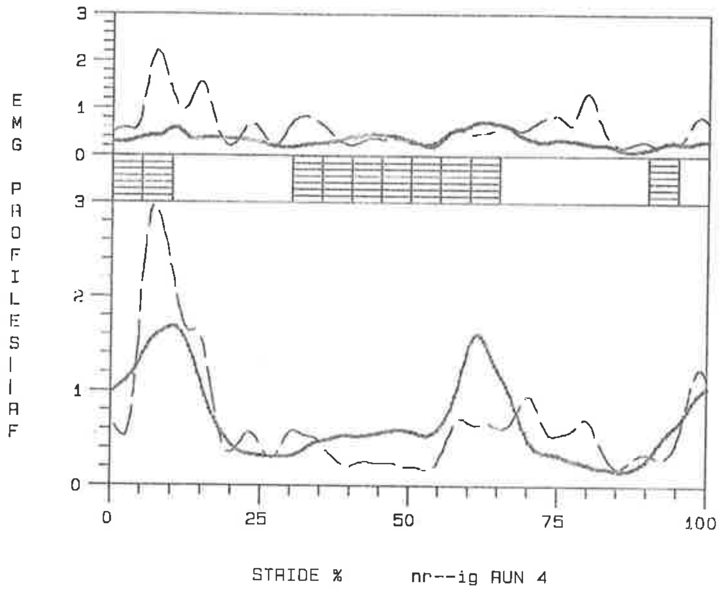

Figure 1. RF EMG profiles for normal and IN repair populations.

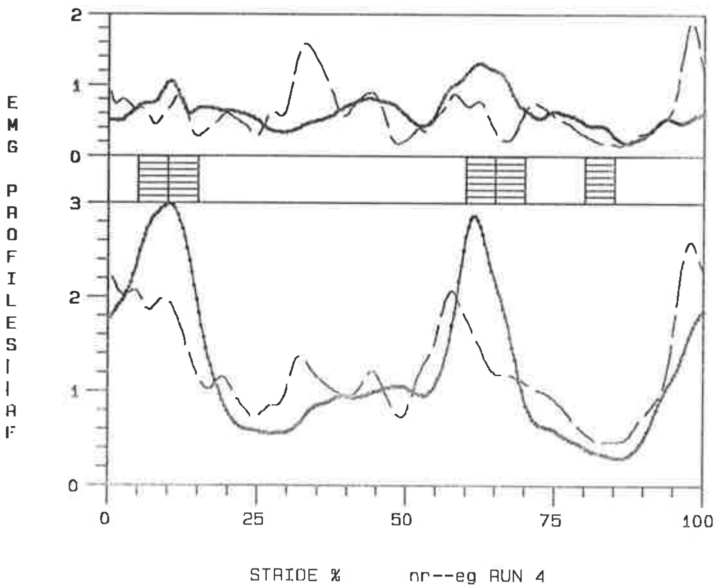

Figure 2. RF EMG profiles for normal and EX repair populations.

that the changes induced by the surgical procedure can be appreciated. The significant differences are summarized in Table 1. The type of difference depended upon the surgical procedure.

In RF the injury causes the second phase of activity to be less intense. The IN repair causes the main phase of activity to be multipeaked and the second phase of activity to occur later. Also the second phase is still less in intensity, Figure 1. (The two sets of profiles being compared are plotted in a figure with three windows. The lower window contains the averages and the upper window, the standard deviations. Periods of the stride in which the profiles are statistically significant are indicated by a column of horizontal bars located in the middle window.) The EX repair causes both phases to have almost equal intensities and advances the occurrence of the second phase. In addition the activity is distributed throughout the stride; there tends to be more activity between the phases, Figure 2.

In VL the injury causes the small second phase to occur later. Both of the techniques make the main phase occur earlier and keep the second phase later. The IN repair also tends to make the main phase multipeaked.

In VM the injury causes the small second phase to occur later or to not exist. The surgical procedures do not change this. However, they make opposite changes in the timing of the main phase, the IN repair causing it to occur later and the EX earlier.

In ST injury causes activity during early swing and this is maintained after surgical repair. The procedures cause a pronounced difference in the main phase of activity, the IN repair causes a reduction before foot strike whereas the EX repair causes the main phase to occur earlier and to be less in duration. They also have opposite effects on the level of activity during midstance.

In BF injury causes the main phase to occur later and the small second phase to occur during early swing instead of during midstance. Surgical repair causes a modification of these changes. The IN repair basically causes the main phase to remain delayed and shortened in duration. The EX repair causes the main phase to occur earlier.

In GS the injury caused the main phase of activity to be shorter in duration and phases of activity during loading and

TABLE 1 - EMG PATTERN CHANGES			
	injury	IN repair	EX repair
RF	P2 less	MP multipeaked P2 less P2 later	MP and P2 equal P2 earlier variable activity
VL	P2 later	MP multipeaked MP earlier P2 later	MP earlier P2 later
VM	P2 later/none	MP later P2 later/none	MP earlier P2 later
ST	ESW activity	P2 in ESW MST less MP less at LSW	MP narrower MP earlier MST, ESW activity
BF	MP later P2 later	MP later MP narrower P2 later	MP earlier ESW activity
GS	MP narrower phase in loading phase in swing	MP narrower phase in loading phase in swing	MP earlier

ESW - early swing
MST - midstance
MP - main phase of activity
P2 - second phase of activity

swing. The IN repair did not change this activity; however, the EX repair caused the additional phases to disappear and the main phase to be activated sooner.

SUMMARY

The muscular activity patterns needed to maintain a functional knee joint depend upon the surgical procedure used to repair the ACL deficient knee. The quadriceps muscles underwent changes in timing of the smaller phase of activity and the main phase tended to be multipeaked in the IN repair group. In the hamstring muscles the main phase of activity tended to occur earlier and swing phase activity developed after EX repair. After IN repair the main phase was delayed or multipeaked and defined phases of activity occurred during swing. The IN repair did not seem to affect the GS whereas the IN repair caused the main phase to occur sooner and the other phases of activity did not appear.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

ACTIVATION PATTERNS OF SHOULDER MUSCLES IN GOAL DIRECTED MOVEMENTS

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INTRODUCTION

Our interest in the shoulder joint (5,6) finds its motivation in the field of rehabilitation concerning shoulder arthrodesis, humerus endoprosthesis, and wheelchair propagation. To gain insight in the function of shoulder muscles, goal directed arm movements were studied, mainly consisting of an anteflexion or retroflexion in the shoulder. Electro-myographic activity was recorded in 13 muscles in the shoulder and the arm.

For fast goal directed movements, the EMG generally has a triphasic shape (2). The first phase causes acceleration and the second phase causes deceleration of the limb. The third phase reduces the duration of movements (3,4).

In the current contribution, the EMG will be considered of : muscles acting on the scapula, muscles acting on the humerus and bi-articular (shoulder-elbow) muscles. The occurence of triphasic patterns will be regarded as an indication of the contribution of shoulder muscles to the acceleration and the deceleration of the limb. A question to be answered is to what extent muscles acting on the scapula contribute to an acceleration or deceleration of the limb or just serve to properly orientate the scapula which can be regarded as the moving base for the glenohumeral joint.

METHODS

Of 8 healthy male subjects, goal directed movements were recorded, see figure 1. This presentation concerns both forward and backward movements with a stepsize of 0.2 m made at maximum velocity. The manipulator simulated a mass of 0.6 kg and a damping of 0.6 Nsm^{-1}.

The EMG was recorded with bipolar surface electrodes consisting of gold-plated brass cores (diameter 7.5 mm) embedded in flexible synthetic material at an interelectrode distance of 21.5 mm. The EMG signal was pre-amplified by a differential pre-amplifier (gain 100, input impedance 100 MOhm, CMRR 86 dB). The reference electrode was fixed on the pre-amplifier which was attached to the waist. After further amplification the signals were stored on FM-tape. Later, the signals were sampled at 1,000 Hz, rectified and averaged over 10 ms.

To enable a quantitative analysis of patterns, an algorithm was developed which scans EMG signals for significant changes in activity. An illustration of switches detected with this algorithm is given by the dotted lines in figure 2. In this figure, a pre-movement antagonist activity can be seen which may be assumed to contribute to a maintainance of the initial position against gravity.

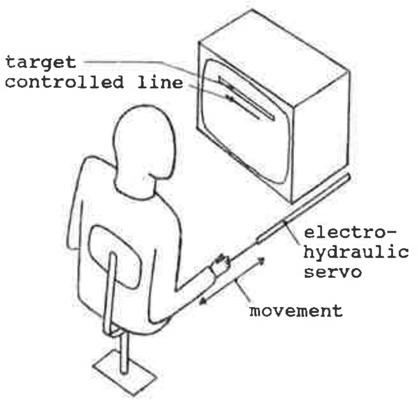

fig. 1. Experimental setup.

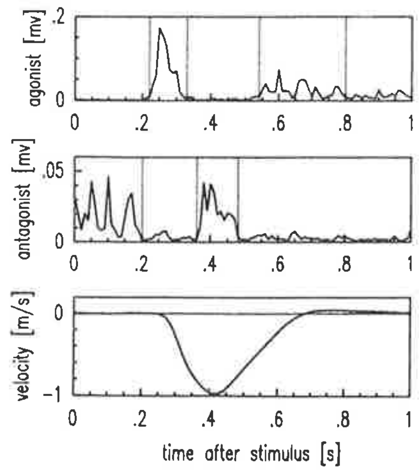

fig. 2. Triphasic pattern in EMG of agonist (post.deltoid) antagonist (ant.deltoid) and velocity trace of a retroflexion movement.

RESULTS AND DISCUSSION

Functions and patterns

In the movements considered, the clearest triphasic patterns were found in the following two muscle groups :

"anteflexors" : the anterior deltoid and the pectoralis major (clavicular part), acting as agonists during anteflexion and mostly as antagonists during retroflexion.

"retroflexors" : the posterior deltoid and the latissimus dorsi, acting as agonists in retroflexion movements and mostly as antagonists in anteflexion movements.

So, these 4 muscles acting on the humerus appear to play an important role in the acceleration and deceleration of the limb.

Of the muscles acting on the scapula, the activity of the three parts of the trapezius and of the serratus was recorded. In most subjects, the trapezius descendens and the serratus show an agonist pattern during anteflexion and the trapezius transversalis and ascendens show an agonist pattern during retroflexion. In the opposite movement direction, an antagonist pattern was observed in only few of these cases. This suggests an important role of muscles acting on the scapula mainly during the acceleration of the arm.

In bi-articular (shoulder and elbow) muscles and in muscles acting only on the elbow, the patterns deviated from the triphasic pattern and patterns were subject dependent.

The latissimus dorsi and the posterior deltoid both act as retroflexors, but the algorithm detected a significant difference in phasing. In 7 out of 8 subjects bursts of the latissimus start and end up to 60 ms before bursts of the posterior deltoid. A possible explanation for this phenomenon is that the latissimus has longer muscle fibres and a longer tendon than the posterior deltoid and therefore (probably) exhibits a slower response of force to activation. Baratta and Solomonow (1) demonstrated a comparable relation of muscle dynamics with muscle architecture. The earlier activation of the latissimus can be seen as a compensation, made in the CNS, for its slower dynamics.

Another difference is that the latissimus has its origin on the trunk and the posterior deltoid on the scapula while both act on the humerus. When considering agonist patterns of muscles acting on the scapula during retroflexion it was found that the onset of the first agonist bursts of the trapezius transversalis and ascendens don't differ significantly from the latissimus dorsi ($p > 0.3$) and preceed the posterior deltoid ($p < 0.001$). This suggests that an activity of the muscles having their origin on the trunk and their insertion on the scapula or the humerus might preceed the activity of muscles having their origin on the scapula and their insertion on the humerus. This would indicate that the rotation of the scapula preceeds the rotation of the humerus, also because of the relatively low innertia of the scapula. However for the anteflexors, such a relation between timing and insertio (trunk or scapula) was not clearly found.

To enable a comparison of agonist and antagonist timing, the average timing of the anterior deltoid and the pectoralis major is further considered as the timing of anteflexors. The average timing of the posterior deltoid and the latissimus dorsi is considered as the timing of the retroflexors. For this average timing, the duration of the first agonist burst ranges from 69-171 ms and the duration of the antagonist burst ranges from 94-174 ms. The start of the first agonist burst comes 22-82 ms after the inhibition of the antagonists. A silent period of 6-72 ms was found between the first agonist burst and the antagonist burst. In some cases a silent period was found between the second and the third burst. In other cases these bursts overlapped in time.

The second agonist burst had lower amplitudes than the first agonist burst. Reduced amplitudes are also found in the antagonist burst compared to the first agonist burst of the same muscle in the opposite movement direction.

Time-optimality

The subjects were instructed to reach the target position as fast as possible. So, a time-optimal control strategy could be expected. Generally, the time-optimal control is a bang-bang control; the inputs switch a number of times between minimal and maximal values. Such a switching between minimal and maximal inputs suggests a switching between maximal agonist and maximal antagonist activity which is in contrast with the experimentally found silent periods and submaximal amplitudes. However, no analytical outcome is available on what is the time-optimal control for a non-linear system like the movement system. Therefore, a non-linear muscle model was simulated (7). An optimization algorithm was used to find the time-optimal input (activation pattern) for this model. Thus, it was found that the time-optimal input (as expected) switches between maximal agonist and maximal antagonist activation. Also an input similar to the experimentally found activation patterns was simulated. This input contains a silent period between the first two bursts, the second and the third burst have submaximal amplitudes (50%). This input led to a "near time-optimal" movement with a duration 9% larger than the minimal duration. This suggests that a criterion is minimized which also takes into account the effort spended.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

SHOULDER MUSCLE EMG IN NORMALS AS A REFERENCE FOR CLINICAL EVALUATION

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INTRODUCTION

In order to use electromyography (EMG) as a clinical tool in for example the analysis of muscle function during voluntary dislocation of the glenohumeral joint, it is necessary to obtain normal EMG patterns in persons without any shoulder complaints for comparison. Important items are then among others the choice of normalisation method of the EMG's and the standardisation of experimental conditions and electrode location as a means of decreasing the effect of anatomical and physiological factors on the EMG signal. Skin thickness, motor point location, muscle strength, muscle moment arm, muscle architecture, muscle force-length relations and muscle force-velocity relations are such important factors in this context.

A specific problem in this EMG study was the fact that shoulder muscles were investigated. The shoulder is a complex system with several polyarticular muscles, and bones with a large range of motion, causing a remarkable length change of the muscles and redirection of muscle force with respect to external forces.

This myriad of difficulties makes it hard to choose a suitable normalisation method. The following focuses on the methods used in order to standardise experimental conditions and to diminish the influence of some of the factors mentioned above. Some results are presented.

METHODS

Twelve muscles (for names see fig.1) of the right shoulder of twelve healthy young subjects without shoulder complaints were investigated. At five specified abduction and anteflexion angles (30, 60, 90, 120 and 150 degrees) isometric EMG's were collected during one second. This routine was carried out twice. The order of elevation angles was chosen randomly. Elevation angles were measured with a goniometer. At every angle also the maximum moment and EMG were measured simultaneously.

The bipolar surface electrodes had a fixed interelectrode distance of 21.5 mm. The pick-up area was made of gold plated copper and was 7.5 mm large. The skin was shaved, lightly rubbed with sandpaper and treated with 70% ethanol. Electrode jelly was used. The electrodes were placed along the bundle direction with adhesive tape. Since most shoulder muscles are wide and divergent the electrodes were placed at a standardised position with respect to bony landmarks.

The myoelectric signal was preamplified (AC coupled, gain 40 dB, input resistance 100 MOhm, CMRR > 85 dB) and then analog high pass filtered (cutoff at 10 Hz, 1st order). Subsequently, the signal was amplified (single ended) with a variable gain per channel up to 60 dB. This factor was used in calculating EMG signals at the skin. The signal was stored on tape (Recorder: RACAL Store 14 D). Before analog to digital conversion (1 KHz, DT 2821, 12 bit) an analog low pass filter was used (8th order, cutoff at 500 Hz). The complete analog system had a noise factor < 1 microvolt. The digital signal was rectified and the offset removed. Mean amplitude (Memg) of one second registration was calculated. This value was used for further analysis.

RESULTS AND DISCUSSION

In order to use EMG as a tool in muscle function analysis in a clinical setting, two conditions must be met. In the first place the EMG pattern should be reproducible within a person, and in the second place, a common pattern with rather small interindividual variations should be present in the normal group to distinguish a patient group from.

As an indication for the reproducibility the linear correlation coefficient between two trials was calculated for every subject. For two subjects this coefficient was also calculated pairwise for measurements on three days. The linear correlation coefficient between repeated measurements on consecutive days and trials was > 0.9 in over 80% of the dynamic measurements, which indicates that the recorded EMG's were highly reproducible within one subject. It is assumed that these numbers are valid for the static measurements as well. As compared with literature (e.g. 3) the present EMG values did not deviate much.

When individual EMG patterns were compared, there appeared to be quite a variety among subjects. These observations lead to the assumption that there is also a variety in the way muscles are used to raise the arm and move the scapula accordingly. Most striking was the fact that for most muscles Memg increased from 30 through 150 degrees humerus elevation, though the net moment to be generated about the shoulder is maximal at 90 degrees. This discrepancy beyond 90 degrees could be caused by decreasing moment arms, muscle length dependency of EMG, shift of motor points with respect to pick up area, or a combination of these factors. In order to decrease these effects the EMG measured at a specific elevation angle (EMGloc) was normalised to the EMG signal collected during maximum voluntary contractions at the same elevation angle (EMGlocmax). In this way a factor Kemg = EMGloc/EMGlocmax was obtained.

In general, normalisation is carried out to quantify the EMG signal and to eliminate factors that influence raw EMG, such as skin resistance and motor point location. In the literature several normalisation methods have been applied (see e.g. 4). Normalisation to EMG measured at some maximum voluntary contraction (MVC) is very common. This method leaves however the influence of position dependent factors such as muscle lever arm, muscle length and distance from electrode to motor point. Our method of measuring and normalisation enables us to reduce this influence of humerus position.

Not only in the raw EMG's but also in the normalised EMG's (Kemg) there is quite a variety in the patterns per subject. This is illustrated by the coefficient of variance of Kemg calculated as standard deviation/mean * 100 % (across subjects). Averaged across muscles this value was 44.2 % (sd 15.4; range 31.2 - 89.2) in abduction and 53.3 % (13.4; 38.7 - 90.1) in anteflexion. Without the biceps this was 40.1 % (7.5; 31.2 - 57.7) in abduction and 49.9 % (7.9; 38.7 - 60.3) in anteflexion. For manual labor tasks of carpenters the interindividual variability was found (1) to be quite large while the intraindividual variability was small.

The factor Kemg increases with increasing arm angle (fig. 1). On average there was for the majority of the muscles an almost linear increase in Kemg during elevation. This was observed in most of the subjects. In static abduction it appeared from mean patterns that deltoideus anterior (DA), trapezius ascendens (TA) and serratus anterior (SA) muscles showed a larger increase than the other muscles (fig.1). For individual patterns this could be confirmed in half the number of subjects for DA and TA and in one third for SA. In the average pattern of static anteflexion the DA, trapezius descendens (TD) and pectoralis pars clavicularis (PCL) showed a larger increase of Kemg than other muscles (fig.1). This was also observed in individual patterns in half the number of subjects.

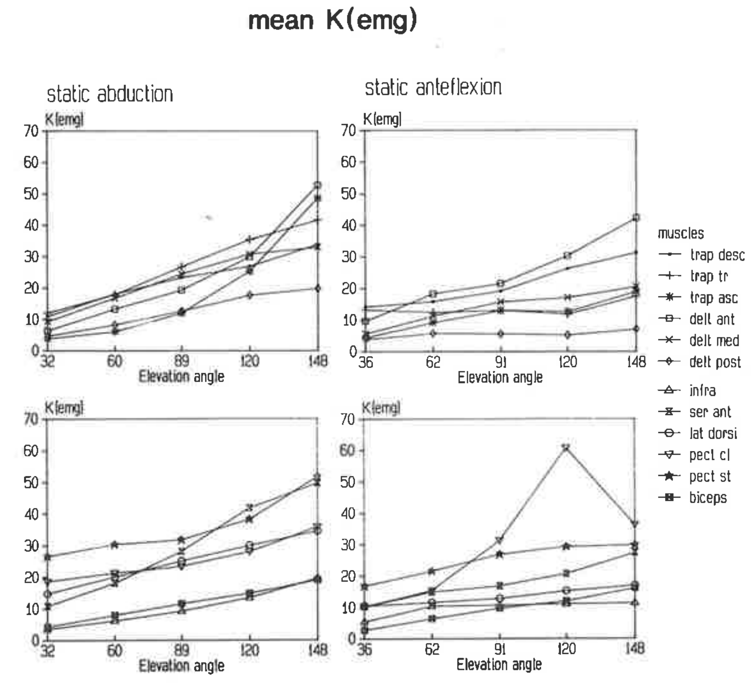

Fig. 1. EMG's of twelve shoulder muscles measured at five elevation angles (EMGloc) normalised to EMG's measured during maximum contractions at the same angle (EMGlocmax).

In interpreting Kemg some remarks should be made. EMGlocmax is dependent on elevation angle, with for most muscles a maximum at 90 or 120 degrees. Some muscles showed a moderate EMGlocmax. Apparently not all muscles are fully used at maximal effort. Kemg can be interpreted as the contribution of the muscle relative to its maximum possible contribution at a specific angle. This relative contribution is not equal for all muscles, probably indicating non-linearities in the (isometric) EMG-force relation or some selection criterion prefering specific muscles. 'Cross-overs' are probably due to increasing moment arms as is the case for DA, and SA (2). From an energy optimalisation point of view it would be most useful to increase the activity of the muscles that have a large moment arm. The generally observed steady increase of Kemg can be caused by the decrease of maximum force caused by change of muscle length, as is indicated by the decrease of the maximal moment around the shoulder joint (not shown).

Through normalisation to EMGlocmax effects of anatomical and physiological factors on the EMG signal have been reduced. It is expected that the influence of changing muscle moment arm and muscle length with elevation angle has been eliminated. One obtains a parameter that describes the relative activity with respect to its maximum at a certain angle.

An application for these normal data may be in the detection of the mechanism that displaces the humerus in luxation patients. The precise cause of luxation is quite obscure but may be sought in muscle discoordination, disturbed propriocepsis and laxity of the glenohumeral capsule. The data from the present study can be used as a reference in the analysis of muscle function in luxation patients. The normalisation method can be applied because the absolute Kemg factors show the relative activation of the muscles at every angle. In the future wire EMG's of the rotator cuff muscles will be added to the data base now available.

CONCLUSION

A fairly large data base of normal muscle activation patterns during humerus elevation has been obtained. The EMG patterns were found to be quite reproducible. Still, the interindividual variability in EMG patterns was remarkable although extreme care has been taken of experimental standardisation and thorough normalisation has been carried out.

We would suggest that the results from the present study could be used in the future in the evaluation of muscle activation patterns of patients. In the conclusions of this patient research one should keep the interindividual variability of normal subjects in mind.

ACKNOWLEDGEMENTS

The authors would like to thank Wim Deegenaars and Hans Frat@man for their technical creativity and Gijs Pronk for general advice.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

NEURAL NETWORK CLASSIFICATION OF MYOELECTRIC SPECTRA

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INTRODUCTION

It has been demonstrated, [1], that the control of a multi-degree of freedom arm prosthesis may be derived from a single channel myoelectric signal (MES). Control decisions in such systems are based on pattern recognition, using a set of features extracted from the MES. To date, attempts at realizing practical control systems based on this technology have had limited success, primarily due to large associated computational loads. Preliminary research by the authors, [2], suggested that neural networks have significant potential for application in the analysis of ME signals.

Previous discrimination tests were made using a set of features extracted using an autoregressive MES model, [2]. By using the first time series parameter, a1, in conjunction with the signal power it was possible to discriminate between different contraction types. Further, it was demonstrated that a perceptron network could be trained to perform this discrimination. Spectral information could be extracted from the raw MES, and would provide a much larger feature set than the time series parameters.

Neural Networks

Lippmann, [3], provides a good introduction to neural networks, and outlines the differences between classification using both single and multi-layer perceptron networks. The structure and training algorithm for single layer perceptrons was developed by Rosenblatt, [4]. These networks are trained in a supervised fashion whereby the inputs and desired outputs are known during the training session. Once the network has been trained, it is then possible to use it for the classification of data not contained in the training set. The structure is characterized by a set of n input, and k output units as shown in Figure 1. The inputs are typically a set of analog values which are scaled by connection weights, W, between each input and output unit.

SPECTRAL ANALYSIS OF MYOELECTRIC SIGNAL

A MES spectral representation was developed in order to provide adequate discrimination between contraction patterns in the upper arm. A fast Fourier transform (FFT) was applied to 500 Hz sampled MES data using non-overlapping record lengths of 32 points, giving a spectral resolution of approximately 16 Hz. In preliminary tests, 125 spectral estimates were averaged in order to reduce the estimation error to approximately 10%. In order that

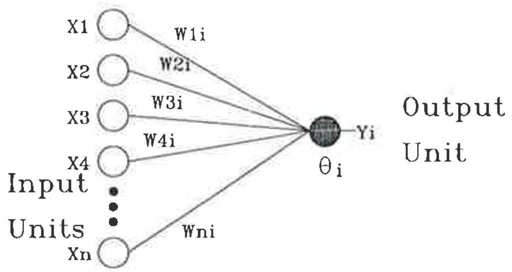

Fig. 1: Neural Network

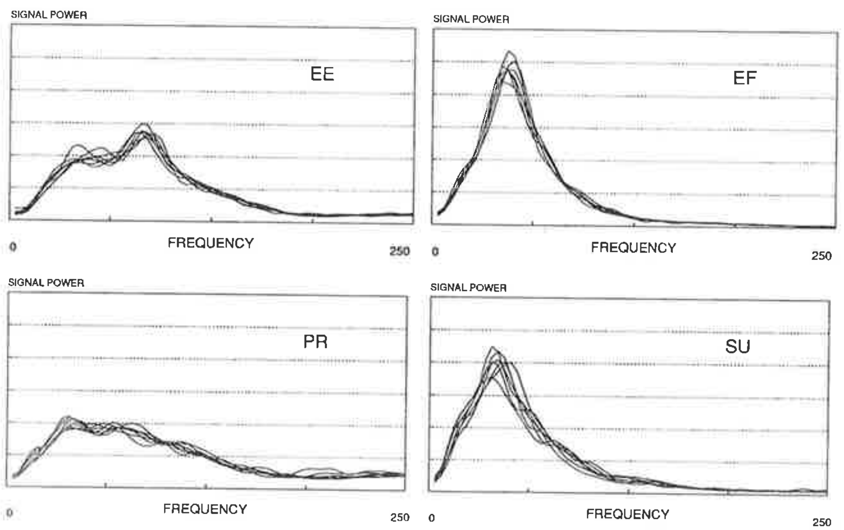

Fig. 2: Spectra for elbow extension (EE), elbow flexion (EF), wrist pronation (PR) and wrist supination (SU)

spectral signatures be independent of contraction level, normalization was performed.

A normally limbed subject was used to gather MES data for each of the four isometric contraction patterns 1)elbow flexion(EF), 2)elbow extension(EE), 3)wrist pronation(PR), and 4)wrist supination(SU). A Beckman silver silver-chloride electrode was placed over each of the biceps brachii and triceps muscles. A series of fifteen sets of 8 second contractions for each of the four functions was recorded. The resulting MES data were digitized and stored for further processing. The resulting spectra are shown in Figure 2 and it is seen that all spectra exhibited characteristic profiles that were consistent for a given contraction type.

CLASSIFICATION USING SPECTRAL SIGNATURES

Classification tests were made using various perceptron network configurations, and employing MES spectral representations as input patterns. A two layer network was first employed to test its ability to discriminate between four functions.

Multiple Layer Perceptron

A two layer (16:8:4) perceptron architecture was trained using Back Propagation, [4], on 300 spectral sets (75 per function). The spectra were estimated from the MES data collected as described earlier except that the sampling rate was varied between 500 and 250 and the number of spectra averaged reduced to 10 or 5. The neural network classifier was then tested on 420 new spectral sets (105 per function). Training and classification results are presented in Table 1.

Trial	Sampling Rate	Number Averages	Training Error (%)	Classification Errors (%)				
				EE	EF	PR	SU	Av.
1.	250	10	4	33	6.7	6.7	20	16.6
2.	250	5	12	50	47	45	20	40.5
3.	500	10	2	29	26	23	18	23.9
4.	500	5	30	-	-	-	-	-

Table 1 Training and Classification Results: Four Functions, Using (16:8:4) Network.

With the network implemented on a Intel 80386 based personal computer, the convergence speed of the network during the training stage was found to be very slow. This prompted an investigation of simpler approaches to the classification problem that would require less computing overhead.

Single Layer Perceptron

The possibility of partitioning the discrimination problem into several linearly separable tasks was considered. In this case single layer perceptrons, with shorter training times, can be used. Three single layer (16:2) perceptrons were tested in order to evaluate their ability to discriminate between spectra by groups. The first groupings are all EE and PR spectra (labelled EEPR), and all EF and SU spectra (labelled EFSU). The first perceptron carries out a discrimination, labelled EEPR/EFSU, between groups EEPR and EFSU. Similarly the other two perceptrons discriminate between the groups EE and PR (labelled EE/PR), and the groups EF and SU (labelled EF/SU).

Again spectra were generated from the MES data collected as described earlier at a sample rate of 500 and number of spectra averaged set to 10. Training and classification data sets for five normally limbed subjects were defined as above, and the results are presented in

Table 2. When compared with the two-layer implementation, the single layer perceptron structure provides comparable or better classification results. Network learning rates were also found to be very rapid, with the most difficult learning achieved typically within less than five minutes on a 80386 based machine.

Subject	Typical Training Error (%)			Average Classification Performance± Standard Deviation (%)				
	EEPR/EFSU	EE/PR	EF/SU	EE	EF	PR	SU	AVG.
A	0	0	14	84±1	83±10	76±2	75±8	79.5±5
B	4	0	0	88±6	95±2	75±4	82±2	84.9±4
C	0	0	4	95±3	93±4	90±1	81±9	89.9±4
D	0	0	3	80±3	87±9	91±1	88±2	86.5±4
E	3	0	0	86±2	94±2	71±5	78±3	82.3±2

Table 2 Training and Classification Results for Five Subjects Using Single Layer Networks

CONCLUSIONS

Spectral representations were developed by ensemble averaging FFT generated amplitude spectra of the MES. It was observed that differences existed between the MES spectral representations associated with natural arm functions. By employing three single layer perceptron structures, and partitioning the classification problem so as to consider pairs of functions, it was possible to develop a system which could be trained efficiently to discriminate among all four functions. No subject training was used in these tests. Based on results from five normally limbed subjects, the average classification accuracy for new spectral data for all four functions was 85 %. It remains to repeat these results for amputees.

ACKNOWLEDGEMENT

This work was supported in part by NSERC grant # A4445.

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EVALUATION OF CONSERVATIVE TREATMENT OF LOW BACK PAIN VIA ELECTROMYOGRAPHY AND PAIN QUANTIFICATION

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INTRODUCTION

Low back pain (LBP) is a major medical complaint which has become a physical, emotional and economic problem of vast proportions. This study develops a technique which measures the effectiveness of conservative treatment of LBP via quantification of change in pain and integrated electromyography (IEMG) data. Osteopathic manipulation therapy is examined in this study (the methodology may be utilized for other conservative treatments). The treatment consists of manual muscle stretching and separation of spinal facets.

MATERIAL AND METHODS

The IEMG data is obtained from bilateral muscle sites at the L3-L4 (lumbar) spinal level, via surface electrodes, during a ten seconds standing isometric resistance of flexion task, as described by Schultz, et al[1]. The electrodes are placed on the skin surface adjacent to the multifidus, longissimus, rectus abdominus and lateral oblique muscles. The IEMG is measured for five external load levels (0-5.0 kg) with one minute rest periods between loads. This differs from several previous LBP studies which only measure EMG for a static, no load, position. Linear regression analysis of the IEMG vs load data is performed and a slope value (i.e. change in IEMG with change in external load) is obtained. The isometric resistance of flexion task is performed by a LBP patient group (n=13) and a control group without LBP (n=12).

Patient selection criteria include: 1) the presence of lumbar pain of muscular origin and 2) the absence of: progressive neurological changes, weakened or loss of reflexes, atrophy of tissues supplied by derivatives of L5-S1 nerve roots, systemic disorders such as tumors, diabetic neuropathy, visceral pathology and spinal fractures.

TABLE I

PAIN QUESTIONNAIRE DATA

Statistical Analyses via Paired T-Tests which examine pre to post treatment session changes in patient group pain data.

Variable	P-Values for Each Session: One	Two	Three	Four
McGill Pain Questionnaire	0.0004	0.009	0.03	NS*
Visual Analog Scale	0.03	0.002	0.003	0.05
Present Pain Intensity	0.05	0.04	0.008	NS
Pain Diagram	0.03	NS	0.03	0.03
Isometric Task Pain Scale	NS	0.008	0.007	NS

* NS = Not Significant

TABLE II

IEMG SLOPE DATA (IEMG VS LOAD)

Statistical analyses via Paired T-Tests which compare initial to final session Total Erector Spinae (Longissimus plus Multifidus) IEMG Slope Data.

	N	Initial Session Mean Value	Final Session Mean Value	P-Value
Patients	13	46.9	36.8	0.03*
Controls	12	63.7	65.8	0.74 (NS)

* Overall treatment effect (initial - final session) is a decrease in IEMG slope.

TABLE III

IEMG FOR INDIVIDUAL LOAD LEVELS

Statistical analyses via Paired T-Tests which examine changes in IEMG data for individual load levels from pre to post treatment session one for patient group and from session one to session two for control group.

Load (kg)	P-Values: Control	Patient
0	0.49 (NS)	0.31 (NS)
1.5	0.99 (NS)	0.72 (NS)
2.5	0.86 (NS)	0.91 (NS)
3.5	0.53 (NS)	0.45 (NS)
5.0	0.46 (NS)	0.11 (NS)

The patient group IEMG and pain data (Short Form McGill Pain Questionnaire, Pain Diagram, Visual Analog Scale and Present Pain Intensity Scale) is obtained before (pre) and after (post) treatment for one to four sessions and the control group IEMG is measured for two sessions. In addition, patient disability data (due to low back pain) is obtained for each session via the Roland-Morris Disability Questionnaire.

RESULTS

Statistical analysis via an Analysis of Variance (ANOVA) of the control group total erector spinae (i.e. right and left side multifidus plus longissimus) IEMG slope data did not show a significant change between sessions (Figure 1). However, the patient group total erector spinae IEMG slope did show a significant treatment effect from pre to post session one (Figure 2). This treatment effect is a decrease ($p \leq 0.02$) in total erector spinae IEMG slope (in figures 1 and 2, one microvolt IEMG equals eight counts).

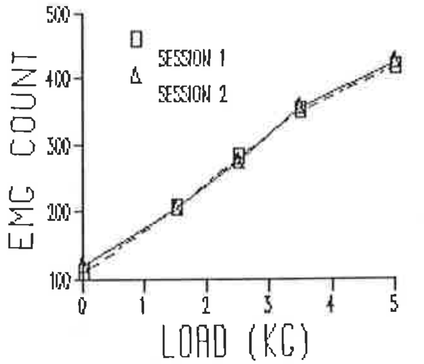

Fig. 1 Control Group. Standing Isometric Resistance of Flexion. Total Erector Spinae slope (EMG vs Load). There is not a significant difference in slope between sessions.

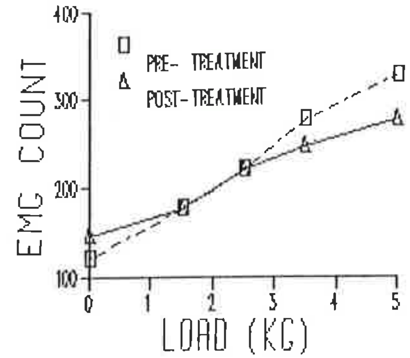

Fig. 2 Patient Group. Standing Isometric Resistance of Flexion. Total Erector Spinae slope (EMG vs Load). There is a significant decrease ($p \leq 0.02$) in slope with treatment for session 1.

There is also a significant decrease in pain data with treatment for each session (Table I). In addition, the decrease in patient IEMG slope (Table II), pain ($p \leq 0.0001$) and disability ($p \leq 0.01$) data continues throughout remaining sessions, as determined by a comparison of initial to final session data.

The IEMG data for individual load levels were statistically analyzed to determine if one can measure a significant treatment effect on LBP by utilizing individual load levels. Previous LBP studies which utilized one load level IEMG data to measure treatment effects have yielded contradictory results. The statistical results showed that one cannot measure a significant change in IEMG for single load levels (Table III).

DISCUSSION

The results suggest that the above described technique is valid for the measurement of the effectiveness of conservative treatment of LBP. The results showed: 1) IEMG slope measurement, which examines IEMG data over several load levels, is a better evaluator of treatment effects on LBP than IEMG for a single load level. However, if individual load levels are used, the higher load level, 5.0 kg, showed the most change in IEMG with treatment ($p=0.1$) and is the preferred weight level; 2) Controls do not show a significant change in IEMG slope between sessions; 3) Patients do show a significant decrease in slope for session one; 4) The decrease in slope continues throughout remaining sessions (not temporary). Therefore, the overall change in slope with treatment (initial to final session) is a significant decrease in slope; 5) When the total erector spinae IEMG slope data is divided into constituent muscle groups, the right and left multifidus muscle groups show the greatest decrease in slope ($p \leq 0.01$ and $p \leq 0.05$, respectively). The multifidus muscle is a cause of several LBP syndromes, including posterior facet syndrome and entrapment of the posterior rami nerve[2]; and 6) Pain and disability show a significant decrease, with treatment, for each session and overall from initial to final session.

ACKNOWLEDGEMENTS

This research was supported by UMDNJ Foundation Grant #8-88 and American Osteopathic Association Grant #88-06-204.

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BASIC MOTION ANALYSIS

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

LINEAR ENVELOPE EMG AS A BIOMECHANICAL VARIABLE IN THE ASSESSMENT OF HUMAN MOVEMENT

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INTRODUCTION

EMG records have been routinely used in the assessment of human movement. Raw EMG's are most commonly used qualitatively to infer a causal relationship between specific muscles and the kinematics of specific movement. Such records give us some insight into the relative activation of agonists and antagonists and their role in controlling the movement. However, such assessments require considerable experience on the part of the assessor who must allow for the delay between the EMG profile and the muscle tension waveform, and also try to infer each muscle's role as a generator or absorber of energy (i.e. concentric vs. eccentric contraction). Also, the raw EMG is not amenable to the quantification of variability (which is important in repetitive movements such as gait) or for pattern recognition comparisons between two conditions (such as normal vs. pathological).

Fortunately there are many ways to process the EMG to yield information regarding the biomechanics of the muscles responsible for a movement. It is desirable to generate a "smooth" signal that follows the rises and falls of the activation pattern; thus a full-wave rectified EMG followed by a smoothing circuit, such as a low pass filter, appears appropriate. Such linear envelope (LE) processing not only produces a smooth analog signal but was seen to "parallel" the muscle force signal (1). This fact is not surprising as it has been shown that the frequency response of a muscle twitch was a critically-damped second-order low-pass system (2). Thus, in choosing a suitable second-order low-pass filter we are in effect modelling the muscle twitch response. Thus the LE is smoothed and phase shifted so that it approximates the relative tension profile of the muscle, at least under isometric conditions. The LE has been used as the input to more sophisticated muscle models to predict muscle tension under widely varying changes of length and velocity (3). Unfortunately, such models are difficult to implement because of the extensive calibrations and detailed anthropometric information required. However, the LE with suitably chosen cut-off frequency yields considerable information to interpret movement. Figure 1 is presented to demonstrate the relationship between the cut-off frequency and the average twitch time for the muscle being monitored. The associated table shows that filter cut-offs from 1.5 to 4 Hz will model twitch times from 106 to 40 ms. In an EMG-driven optimization model (3) the filter cut-off frequencies were found to vary from 1.2

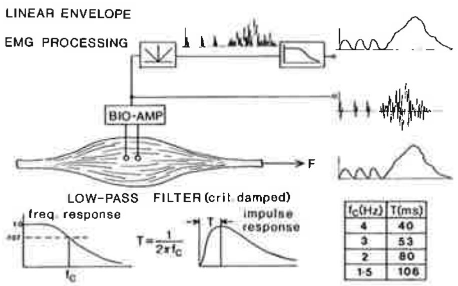

Fig. 1 LE processing of EMG

Hz for the soleus to 2.8 Hz for the rectus femoris. The purpose of this paper is to add to the information already published and demonstrate how the LE aids in the interpretation of human gait.

Relationship Between LE and Joint Moments-of-Force

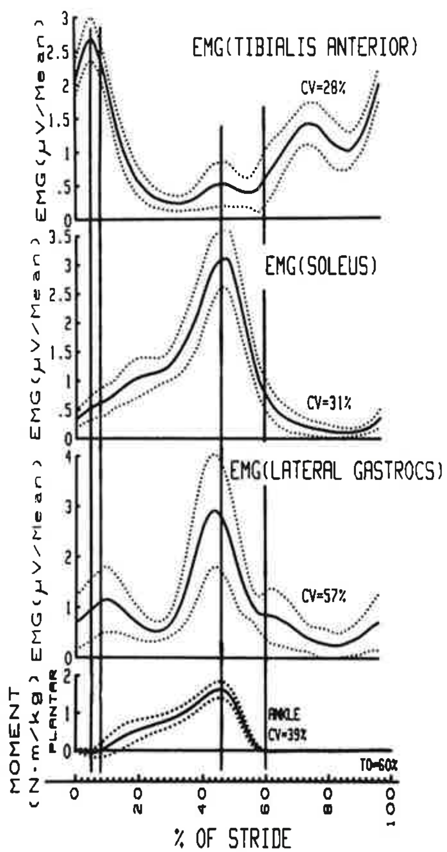

Fig. 2 Ankle moment and LE of 3 ankle muscles

Figure 2 presents the LE from 3 major ankle muscles along with the ankle moment of force. At 5% of the gait cycle there is a net small dorsiflexor moment (to control the lowering of the foot to the ground). The agonist tibialis anterior LE is seen to peak at that time but with a small and increasing co-contraction from the soleus and gastrocnemius. The ankle moment reverses at 8% of stride (TA is seen to decrease as LG and SOL increase). Then the ankle plantarflexor moment increases during mid-stance to reach a peak at push-off within 2% of the time that the SOL and LG reach their peaks. Then as toe-off (TO) is reached (60% stride) the ankle moment decreases to zero in spite of decreasing but significant activity from SOL and LG; the rising TA activity indicates a second co-contraction to stabilize the ankle during the critical transition at TO. In a similar manner we could

demonstrate the use of LE profiles of the quadriceps, hamstrings, gastrocs and gluteus maximus to explain the moments at the knee and hip during walking.

Combining Muscle Velocity With LE Profiles

The LE can be modified with additional information if it is readily available. One of the more important kinematic parameters that would greatly enhance its interpretation is whether the muscle is shortening or lengthening and at what velocity. If the kinematics of the limb segments are known then the origin-to-insertion length of each muscle (in resting lengths, l_0) can be estimated over time (4). If corrections for pennation angle are made the velocity of shortening or lengthening of the muscle fibres, v_f, can be calculated (5). Then the LE profile of each muscle can be shaded according to the polarity and magnitude of the velocity. For near-isometric contractions (v_f less than .25 l_o/s). For low velocity shortening ($.25 < v_f < 1.5\ l_o/s$) the shading was , for fast velocity shortening ($v_f > 1.5\ l_o/s$) the shading was . For lengthening velocities the limits were the same but the slope of the shading was negative. Thus positive slope shading represents positive work by the muscle, negative slope shading indicates negative work done by the muscle. Figures 3(a) and 3(b) illustrate the results of such a velocity analysis and superimposed computer graphics for the soleus muscle. From HC to 8% of stride the soleus shortens as its actvity starts to build up; as a co-

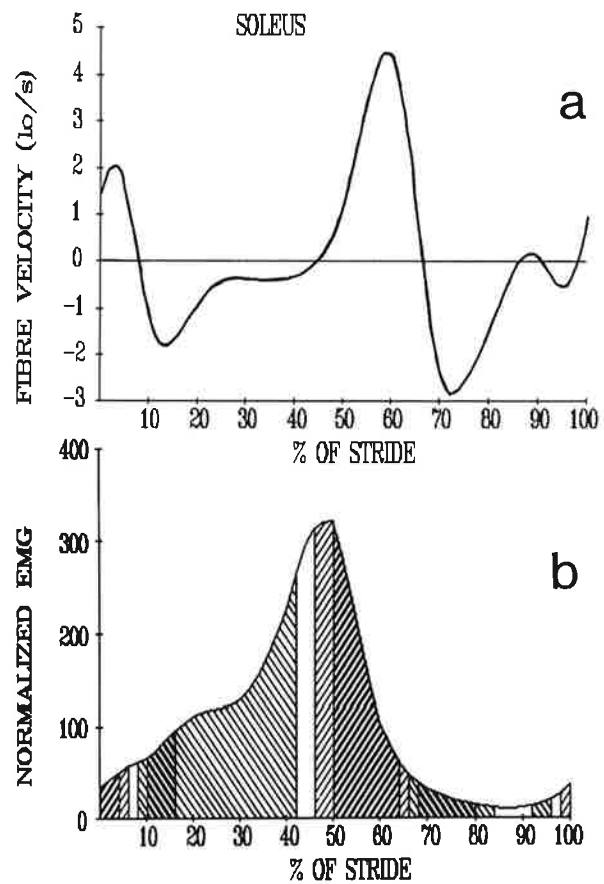

Fig.3(a) Soleus velocity, (b) LE shaded for velocity contraction to the TA which acts to control the lowering of the foot to the ground. Then until 44% stride the soleus lengthens doing negative work to control the forward rotation of the leg over the foot. At 44% soleus activity increases towards its maximum to cause rapid plantarflexion; this push-off phase is the most important energy generation phase in gait. Then shortly after TO (62% of stride) the soleus lengthens in a co-contraction as TA rapidly dorsiflexes to achieve a safe toe clearance.

Ensemble Averaging of LE Profiles

It is critical to know how variable an EMG is from stride-to-stride and how variable inter-subject averages are when used in data bases. Over the past 8 years we have been collecting data on 25 muscles using surface electromyography. Stride-to-stride ensemble averages of the LE for 16 of these lower extremity muscles have been reported (6) along with variability measures. The average variability for the LE profiles over the stride has been documented with a modified coefficient of variation (CV) which is, in effect, a variability to signal ratio. For the 16 muscles reported these stride-to-stride CV's varied from 30% for the more distal muscles up to 60% for the more proximal muscles. Inter-subject ensemble averages were, as expected, seen to have considerably larger CV's and this was largely due to amplitude differences of the EMG's from each subject. When these amplitude differences were removed using an amplitude normalization technique the CV scores for each inter-subject ensemble were drastically reduced to the range seen for intra-subject averages. Pathological gait profiles can be readily compared with these inter-subject averages for a diagnosis (7).

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

NEURAL AND MECHANICAL CONTROL OF A RANGE OF RAPID MOVEMENTS

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INTRODUCTION

Fast movements are accomplished by a multiphasic pattern of muscle activity (1, 2) with at least three reciprocal bursts centrally programmed. The pattern is characterized by a burst of agonist activity corresponding to the accelerating torque, followed by a burst of antagonist activity (decelerating torque) and a second burst of agonist activity (3, 4).

The literature contains many studies comparing movements of different amplitudes, on the basis of electromyograms or kinematics. The results are somewhat conflicting and highly dependent upon movement instructions. With rapid movements, (performed "as fast as possible") investigators generally find patterns with fixed timing and scaling of velocity, acceleration or agonist EMG (5, 6). In contrast, antagonist activity, with these instructions, appears to be unrelated to movement amplitude (7, 8). Recently, Gottlieb, Corcos and Agarwal (9, 10) have proposed organizing principles that resolve some of the discrepancies between previous studies. They propose two different "strategies", the choice between them depending on whether or not speed (and/or movement time) is part of the task specification.

Employing a consistent and unambiguous instruction, we tested the existence of a single strategy, over a full range of amplitudes tested at the smallest reproducible increments. Moreover, we studied movements spanning the full range over which the mechanics of the hand are comparable.

MATERIALS AND METHODS

Movement Task Description

Subjects performed wrist flexions of various amplitudes, while viewing a display consisting of a wrist position cursor and a narrow target. Subjects began each trial in a wrist position close to anatomical position and were instructed to move as

Fig. 1. Increases in movement amplitude are accomplished by scaling pulse heights while pulse width remains constant.

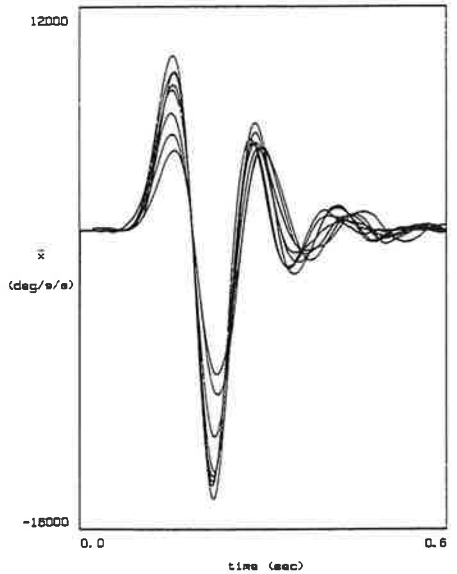
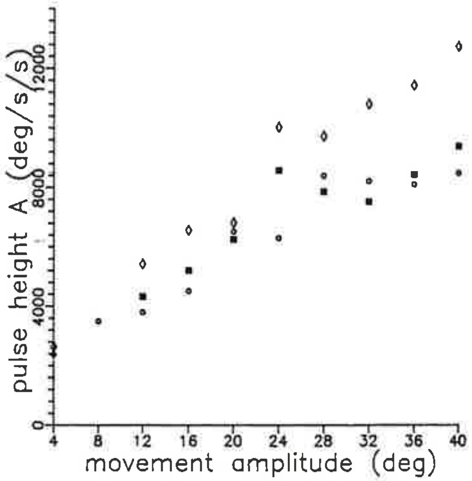


Fig.2. Pulse height A increases monotonically with movement amplitude.



quickly as possible to the target. After each trial, they were informed of the maximum velocity produced, and attempted to maximize this score. The movement amplitude varied from four to forty degrees, in four degree increments. Subjects practiced movements at each amplitude until they could make consistent movements (about twenty trials). Once proficiency was achieved, twenty trials were recorded at each amplitude.

Data Recording and Analysis

Position, and surface EMG's from the flexor carpi radialis and extensor carpi ulnaris were sampled every 0.002 seconds. After each recording session, the ten trials with the most similar dynamics of the twenty recorded for each amplitude were chosen for further analysis. Similarity was assured by graphically overlaying synchronized traces. To avoid filtering artifacts and the lag inherent in our analog acceleration signal, velocity and acceleration were derived digitally. An identified model of wrist mechanics was used to deduce net muscle torque (7) from acceleration.

RESULTS AND DISCUSSION

Movement amplitude was modulated by changes in peak torques with the timing of the torque trajectory remaining nearly constant (Figure 1). Maximum initial torque (pulse height A in Figure 2) increases monotonically with movement amplitude over most of the range, and appears to be the controlling factor driving movement amplitude. Decelerating torque impulse (pulse height B in Figure 3) scales directly with pulse height A, and the slope of this relationship is identical for three subjects. The integrated deceleration pulse (pulse area B) also increases with amplitude of movement. Movement amplitude does not appear to be modulated by pulse widths.

The scaling of initial acceleration is accomplished by scaling initial agonist EMG amplitude. Total integrated agonist EMG activity increases monotonically with movement amplitude (Figure 4). Scaling of acceleration peak slope as well as the peak slope of the initial agonist burst with movement amplitude is also observed. A similar scaling does not occur for antagonist EMG activity: integrated antagonist EMG, if anything, declines with increasing amplitude.

The modulation of torque with relatively fixed timing that we measured is consistent with the "speed-sensitive" strategy defined by Corcos et al. 1989 (8), applied to the agonist motoneuron pools. However, our results suggest an even more economical strategy for the antagonist pools: excitation independent of movement amplitude, with the appropriate scaling of antagonist torque arising from the mechanics of the stretching muscles.

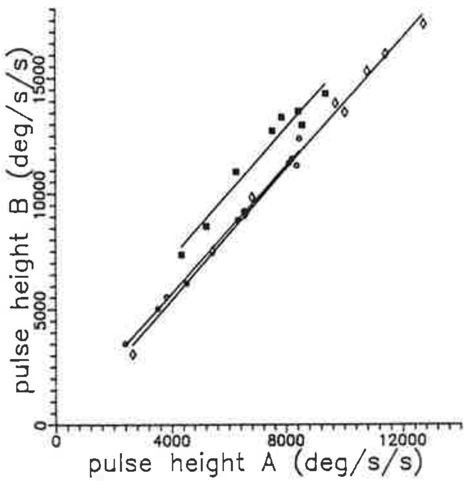


Fig. 3. Pulse height B scales directly with pulse height A.

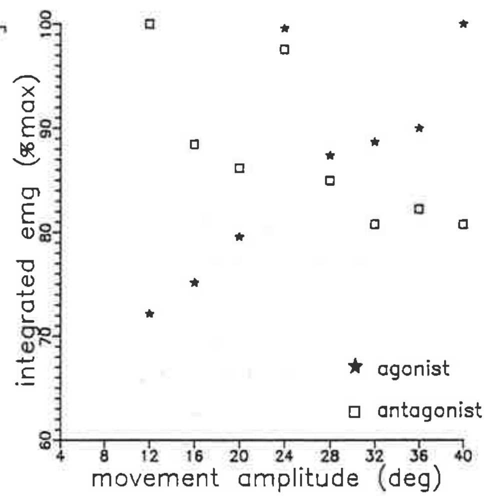


Fig. 4. Integrated agonist EMG increases with movement amplitude while antagonist activity, if anything, decreases.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

THE EVALUATION OF HUMAN JOINT LOAD ESTIMATION METHODS THROUGH A 2-D PENDULUM MODEL

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INTRODUCTION

The joint loads that control the limb and body motion are defined as total forces and moments acting at that joint. They are produced by the muscles and passive soft tissues around the joint. Since in most situations one can not directly instrument any joint of interest to measure the total loads in that joint, the joint loads have to be estimated. Usually, the joint loads are estimated by substituting the measured displacements of the links and their estimated time derivatives into Newton's equations of rigid body motion. The amplification of the original noise due to the differentiation process makes the joint load estimates questionable at best [1]. This study aims to evaluate the joint force estimation based on different approaches by testing a well controlled, instrumented two degree of freedom swinging pendulum.

METHODOLOGY

The two degree of freedom instrumented pendulum (shown in Fig. 1) used a set of strain gauges to directly measure the lumped joint forces produced by the pendulum's motion. The pendulum was also instrumented at its center of gravity with an integrated kinematic segment which included an array of infra-red light emitting diodes (IREDs) for position measurement monitored by the WATSMART optoelectronic system and a miniature triaxial accelerometer for direct linear acceleration measurement. The analog data from the strain gauges and from the accelerometer was amplified, digitized at $100Hz$

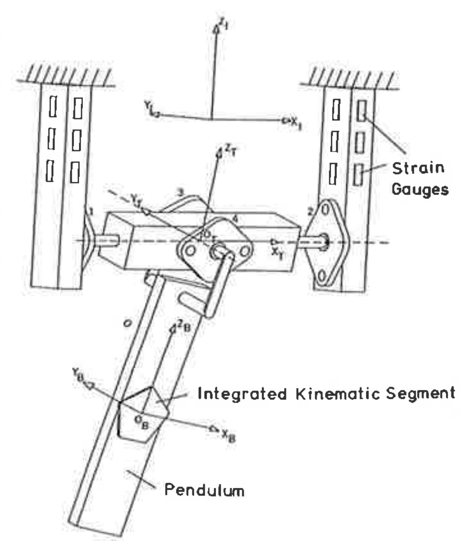

Figure 1: The instrumented pendulum

sampling rate, and synchronized with the WATSMART position measurement through a 16 channel A/D converter (WATSCOPE).

Two coordinate systems were used in the computation process: the inertial reference system **IRS** which is attached to the vertical supports and the body-fixed coordinate system **BCS** moving with the pendulum. The joint forces were measured directly in **IRS** by the strain gauge transducers and also estimated through the inverse dynamics method. The kinematic variables were obtained by two different approaches: the differentiation method and the integrated method. The differentiation method is based on measurement of the linear displacement of the center of mass (C.M.) (in **IRS**) and taking their time derivatives to obtain the linear accelerations. The integrated method is based on direct measurement of the linear displacement and acceleration (in **BCS**) at C.M.. Since DC accelerometers are sensitive to the gravitational acceleration, the effect of gravity g should be corrected from the accelerometer's output ($\mathbf{a}_{acc}$) to obtain the true acceleration ($\mathbf{a}_{cm}$) as shown in equation (1):

$$\mathbf{a}_{cm} = \mathbf{a}_{acc} - T_{BI}\mathbf{g} \tag{1}$$

where T_{ij} is the rotational transformation matrix from coordinate system j to i. T_{ij} was computed through a rigid body kinematics analysis by the TRACK© software package (developed by M.I.T.) that extracted the six degrees of freedom of rigid body motion based on the linear displacement measurements.

The kinematics obtained by both approaches were substituted into the Newton-Euler equation of rigid body motion (equation (2)) to derive the joint force estimates (F):

$$\mathbf{F} = mT_{IB}\mathbf{a}_{cm} - m\mathbf{g} \tag{2}$$

where m is the pendulum's mass.

RESULTS AND DISCUSSION

The measured and the estimated joint forces are shown in Fig. 2. The X, Y and Z components of the joint force and its total magnitude as measured by the strain gauges are compared to the joint force estimates by the integrated method and by differentiation method with $5Hz$ and $15Hz$ low pass filters. The estimated joint forces based on the integrated method are practically on top of the measured force trajectories. The error in the estimates was less than 2%, creating a reliable and accurate assessment of the actual joint forces. The

force estimates by the differentiation method, however, were highly dependent on the filtering scheme. The $5Hz$ low pass filter generates much better estimates of the joint forces than the $15Hz$ one, especially close to both ends of the time window. The magnitude of the total force shows good correspondence between the measured and the estimated forces by the integrated method. The best overall correspondence of force estimates based on the differentiation method is obtained only by applying the $5Hz$ low pass filter.

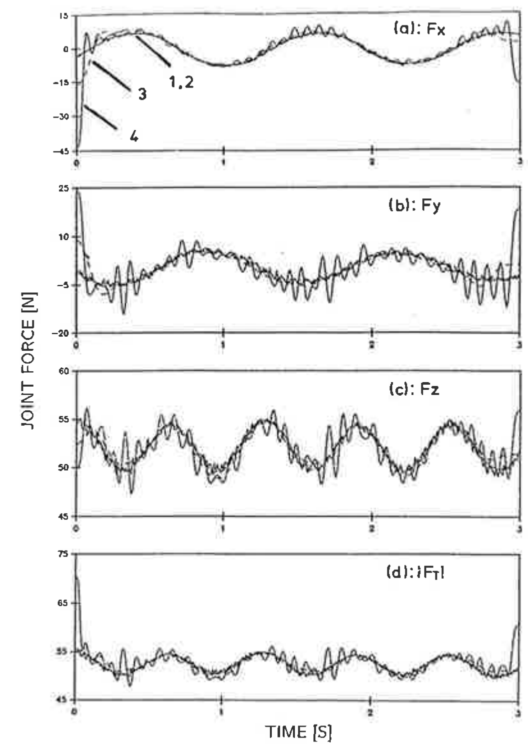

Figure 2: The joint forces during pendulum's motion. 1 - force measured by strain gauges; 2 - force by the integrated method; 3, 4 - forces by differentiation method with low pass filter at $5Hz$ and $15Hz$

The spectrum analysis of the measured joint forces shown in Fig. 3 clearly reveals that in addition to the basic harmonic oscillation at a frequency of $0.78Hz$, there is also a high frequency oscillation at about $12Hz$. The estimated forces by the integrated method followed not only the fundamental oscillation but also the high frequency vibration (see Fig. 3). It is expected that the differentiating method whose low pass filter has a cut off frequency of less than $10Hz$ may be able to capture the basic component, yet it will lose the high frequency band information. With higher cut off frequency, the signal to noise ratio can not be improved and therefore, the output creates significant error in the predicted force.

The close correspondence of the estimated joint forces from integrated method to the measured ones suggested that such an approach of combining position and acceleration measurements to derive high quality joint force estimates is more accurate than the differentiation method, and much simpler than multi-accelerometer measurement schemes [2] and error models [3].

CONCLUSION

The instrumented pendulum was designed to evaluate two joint force estimation methods that are based on the solution of the 'inverse dynamics problem': the integrated kinematic segment method and the differentiation method. By integrating the three elements, namely accurate position measurements, rigid body kinematic analysis that extracts the six degrees of spatial motion of a rigid body, and the use of this information to dynamically calibrate the accelerometer's output one can obtain high-quality 'non-invasive' joint force estimates.

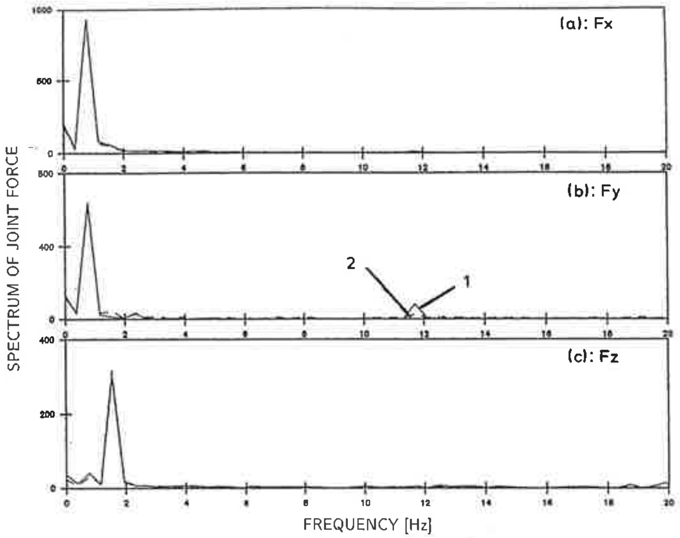

Figure 3: The spectrums of joint forces. 1 - force measured by strain gauges; 2 - force by the integrated method

ACKNOWLEDGEMENT

This research was supported by a grant from the Whitaker Foundation and by National Science Foundation grant NO. EET-8809060.

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Electromyographical Kinesiology
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ELABORATION OF EMG DATA FOR THE MULTIFACTORIAL ANALYSIS OF HUMAN LOCOMOTION

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INTRODUCTION

The multifactorial analysis of human locomotion is a new and powerful approach to study the motor control mechanisms and quantify functional modifications (loss, recovery).
The usual measurements include: kinematics of the human body movements, ground reaction forces, joint moments and powers and EMG activity.
A basic step for a clinical application of multifactorial analysis is to obtain a restricted number of parameters to describe the movement. This task is very complex especially for the analysis of EMG data.
The aim of this study is to describe, through a pattern recognition algorithm, the motor activation behaviour of the main ankle muscles during locomotion.

MATERIAL AND METHOD

Time and amplitude normalisation of surface EMG signal is the first step to get comparable data from different trials and subjects. A good parameter for amplitude normalisation is the mean value of the rectified signal computed over a full stride (Shiavi,1987).
Averaged data of different subjects and muscles have been widely used for the description and statistical comparison of motor strategies (Winter,1987; Shiavi,1987)
A different approach to the analysis of EMG signals is based on pattern recognition techniques such as Karhunen-Loewe expansion (KLE) (Patla,1985) that makes it possible to describe a set of time functions as a sum of features.
In line with the aim of pattern recognition, the KLE technique extracts a number of features which can reconstruct the original data with a reduced loss of information. The analysis of a set of these features should ensure an easier interpretation and comparison of EMG data .

TABLE I
Eigenvalues of the firt five features of each muscle

Features		TA	SO	GaM	GaL
	112	49.43	47.21	53.03	50.97
1	132	51.55	54.44	57.37	54.69
	RUN	55.36	57.30	58.66	59.06
	112	11.20	9.22	10.82	17.00
2	132	6.86	5.73	5.44	12.46
	RUN	4.17	6.03	5.61	4.99
	112	6.11	8.53	3.55	5.59
3	132	4.83	5.23	4.95	3.93
	RUN	4.07	3.72	5.23	4.97
	112	3.39	7.42	3.21	3.23
4	132	2.99	3.78	3.97	3.67
	RUN	3.41	3.65	3.35	3.35
	112	3.38	4.00	3.20	2.86
5	132	2.98	3.07	3.63	3.61
	RUN	3.40	2.87	2.70	3.14

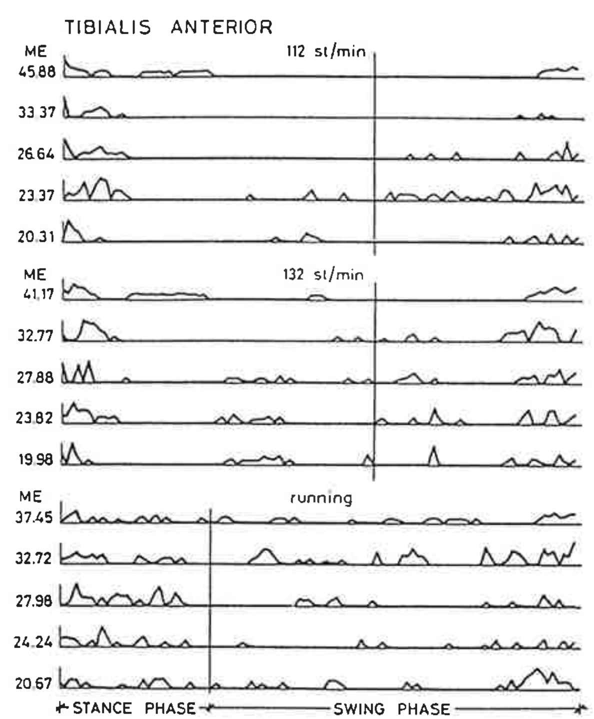

Fig. 1. Five features of TA during walking and running

The features obtained by means of KLE algorithm are the eigenvectors of the autocorrelation matrix of the original data (Fukunaga,Kontz,1970) .
As the eigenvectors are orthonormal, the features are independent of one another and each eigenvalue is an index of the variance of the feature. The choice of the number of eigenvectors to be used in describing the signal can be made by setting a threshold value for the eigenvalues.
A cumulative mean square error (ME) has been defined by Patla (1985)

$$ME(k) = (\Sigma_{1,k} \lambda_i \; / \; \Sigma_{1,j} \lambda_i) * 100$$

where

- k = number of used features
- j = number of original signals
- λ = eigenvalue

ME can be seen as the error of reconstruction the algorithm gives when using k features.
In this study KLE has been implemented and used to analyse the activity of four muscles acting on the calf joint (Ta, GaM, GaL, So) recorded during walking at the cadence of 112 and 132 steps/min and running (average speed 3.5 m/s) on ten male subjects. The signals, were recorded with surface electrodes and a telemetric system at 500 Hz sampling rate.
Before the KLE application the EMG of each trial was normalised during time (cycle of locomotion), reduced to a set of 100 values, rectified and normalised with the average value.
Mean values and standard deviations of omogeneus muscles and locomotion condition have been computed and used to evaluate the consitency of our data with those reported in literature.

RESULTS

Table I shows the eigenvalues of the first five features of each muscle and condition for our population. In all the conditions the content of information associated with these features ranges from 69 to 80 % of the whole signal and more than 50% is covered by the first one. The remaining content of information is almost uniformly distributed over the following 95 computed features.
As the eigenvalues and the correlated ME are indexes of the signal variance, in line with the results reported by Shiavi, it is possible to observe a higher variability during slow walking pointed out by the lowest eigenvalues of the first feature and by the highest number of features needed to obtain the same ME.

Running, on the other hand, is characterised by the lowest variability.
When the behaviour of each muscle is analysed, the Tibialis Anterior shows the greatest variability in all the conditions. This phenomenon could be related to its function of corrector of the perturbations arising during the phase of contact with the ground. Figure 1 shows the first five features of TA.
These data have been used to identify the different strategies implemented to perform the required movements.

Tibialis Anterior During walking the well known TA pattern of activation is confirmed by the presence of significant features before and after the contact of the foot, while during running the activation spreads also during the late stance and swing phase. The main difference between the two walking conditions is the arising of significative bursts of activity during the second half of stance and early swing, when the cadence increase.

Gastrocnemius Medialis Fast walking is characterised by the disappearing of activity in the swing phase and by the reduction of amplitude during stance. Running determines a higher activity during the stance phase and the first half of the swing.

Gastrocnemius Lateralis There are no significative differences between the two walking cadences and in both conditions there is no activity during the swing phase. During running there is a higher activity during stance while during swing the activity appears only from the third feature (low statistical significance).

Soleus During fast walking there is more activity in the swing phase and no important differences in the stance phase. During running the activity generally increases in the stance and swing phase.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

MOVEMENT ANALYSIS FOR CLINICAL PRACTICE. AN ATTEMPT TO BRIDGE THE GAP: THE CAMARC PROJECT.

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INTRODUCTION

Movement Analysis (MA) is the set of methods and techniques aimed at a quantitative assessment of human movement. In addition to the clinical observations, the MA process involves the objective measurements of movement, muscle action, forces and energy consumption. Typical measured variables are the position of body landmarks, electromyographic signals, forces exchanged with the environment and oxygen consumption.

The collected data are processed by computational procedures that, based on an appropriate model of the moving body (Neuro-Musculo-Skeletal model), allow the calculation/estimation of physical variables not directly measurable.

If the process involved the mere collection of the various measurements and the suitable storage and visualization of the large quantity of measured and calculated data, MA would have little value. All data must be correlated, interpreted and applied to the specific patient, or class of homogeneous patients, in order to attain the functional assessment of a motor disorder. This will allow, at the same time, the growth of a Clinical Knowledge (CK) based on MA findings.

MA has a long history, that, in terms of the application of the scientific method, dates from Borelli's work (1679). Milestones in such a history were the advent of photography and the subsequent adoption of stereo-photogrammetric techniques, the assumption of the stereotype of the "rigid body", the development of instruments for measuring ground reaction forces. Also the technical findings relative to the EMG capturing and processing gave a strong contribution to MA. Internal forces, e.g. muscle and tendon tensions, are not measurable through non-invasive methods. They have to be estimated by means of suitable NMS models, as the mechanical energy of body segments .

The above modelling approach implies the assumption of simplifying hypotheses on the mechanical characters and the behaviour of the human body. Relaxation of some hypotheses always lies on the use of more computation and implies harder constraints on the accuracy of the measured data.

The achievements of Biomechanics and recent developments in instrumentation make MA mature for the application in fields ranging from Kinesiology to Ergonomy, Sport Medicine and obviously Rehabilitation. In this latter field, at least in Europe, MA has received limited clinical acceptance .

In the following the main reasons for such a gap among research findings and clinical applications will be mentioned. A short outline of the CAMARC project, which is aimed at bridging this gap, will be drawn. Finally the achievements and perspectives of the project will be outlined.

THE REASONS OF THE GAP

Factors hindering the clinical acceptance of MA can be summarised as follows (1) :

1. Misunderstandigs about the application domain of MA: it is mainly a tool for functional movement assessment, usually in already diagnosed motor diseases. MA can be mainly a tool for clinical decision making and for monitoring the effects of treatments, while recent works suggest its diagnostic application in connection with proper Artificial Intelligence (AI) methods (2, 3).

2. Lack of (and difficulties in building) an accredited Knowledge- Base (KB) of quantitative MA: results obtained in individual laboratories are poorly communicable to others. The lack of standardisation of clinical protocols has hampered the coalescence of findings into coherent data and agreed KB (4). The attempts to standardize and to imbed the existing fragmented but multifactorial information into suitable KB are seen as valuable tools for bridging the gap between research and clinical practice (5).

3. Perplexities about the reliability of MA in managing relevant and intrinsic measurement inaccuracies: some questions concern skin motion artifacts (6), the assumption of ideal joints, the accuracy of measurement equipments and its reflexes on the estimated/calculated quantities, the reliability vs. measuring difficulties aspects of body mass distribution estimation, the indeterminacy problems associated with co-contracting, single and multiple joint muscles (7). Relevant findings are continuously appearing in the literature and two conditions should be met for their adoption into clinical practice. First, clinicians should become more familiar with MA methodology and with the basic principles of Biomechanics (8). Second, Users Friendly Interfaces (UFI) should be provided to the practitioners in order to assist them in navigating the large sea of numerical results and to remove their reluctance to rely on MA methods and results (9, 4, 10, 11).

4. Claims against the validity of MA for the assessment of impairments and concomitant disabilities: the current lack of a satisfactory theory of motor ability is matched, in these objections, with the observation that MA seems more apt to assess the impairment than the disability (12). Such fundamental criticism relies on the large distance existing between the current motor tasks performed into the laboratory and the more complex, everyday tasks, with respect to which the disability should be properly assessed. The development of portable instrumentation for long term monitoring of the motor behaviour and of more flexible instrumentation to allow motor performance assessment in a more realistic setting, can partially answer this basic challenge.

THE CAMARC PROJECT

The above considerations led a consortium of academic, public-health and industrial European entities to the development of the CAMARC (Computer Aided Movement Analysis in a Rehabilitation Context) project. It is a precompetitive and prenormative research project supported by the European Community under the AIM (Advanced Informatics in Medicine) programme. Partners in the project are:

- Strathclyde University, Prof. J.P. Paul, Glasgow, UK
- Biomechanics Consultant, Dr.ir. H.J. Woltring, Eindhoven, NL
- University of Pisa, Prof. A. Starita, Pisa, I (Polysens S.p.A. as subcontractor)
- Istituto Superiore di Sanita', Dr.Ing. V. Macellari, Roma, I
- Istituto di Fisiologia Clinica CNR, Dr.R.Bedini, Pisa, I
- Log.In s.r.l., Ing. G. Bianchi, Roma, I
- University of Ancona, Prof. T. Leo, Ancona, I (Main Contractor)
- INSERM U103, Prof. P. Rabischong, Montpellier, F
- Oxford Metrics Ltd, Dr. J.R.W. Morris, Oxford, UK.

The overall objectives of the CAMARC project are:

- the integration of the existing instrumentation by means of suitable hardware and software interfaces ;
- the definition of a comprehensive KB of the assessed MA experience and its implementation by means of suitable Knowledge Representation methods;
- the definition, on the largest consensus basis, of protocols for data capturing and processing, comprehensive of the development of suitable instrumentation, tailored for various clinical applications;
- the constitution of consistent KB using quantitative MA data with the possibility of making them accessible by suitable networking;
- the assessment of suitable UFI aimed at the easy communication and understanding of the results between the users and at the growth of a pertinent CK using quantitative data.

During the current exploratory phase of the AIM programme (June 1989- December 1990), only preliminary results and feasibility studies are being carried out .

ACHIEVEMENTS AND PERSPECTIVES OF CAMARC

The results attained up to now can be subdivided in the following six basic categories:

1. The assessment of existing biomedical MA Knowledge: at present a strong causal model of the human gait has been defined and implemented through the KADS (Knowledge Acquisition and Documentation Structuring) methodology and by means of the NEXPERT^{TM} shell. It is aimed at determining the interactions among the processes forming locomotion and, after that, at providing reliable reasonings. The suitable classification methods will be chosen afterwards, when well assessed kernels of Knowledge and training sets will be built-up.

2. The standardization of test protocols: current Clinical Protocols have been examined in deep, with special attention for those of dr. Gage's and prof. Sutherland's groups. A workshop has been organised on this topic (13). A proposal for the presentation of data for Rehabilitation purposes has been defined, relative to the Neuro-Musculo-Skeletal model for the leg in locomotion and to the definition of the position and attitude of its segments in 3-D. A proposal of standardization of the evaluation tests on Biokinematic Measurement Systems is being developed through the concertation of almost all the manufacturers of such devices.

3. The development of a UFI prototype: the proposal of the Interface is in press (11). The prototype is addressed to the evaluation of human postural behaviour. Another UFI addressed to Gait Analysis is under development.

4. The assessment and the implementation of relevant DSP algorithms: a comprehensive assessment of the most promising algorithms referred to into the literature and the development of particular new algorithms has been undertaken, under CAMARC, with respect to Analytical Body Segment Photogrammetry and Filtering techniques. Particular attention has been also given to the methods for normalization, feature extraction and classification from the classical statistical ones to the Neural Network applications. These methods ideally close the circle with the KB .

5. The analysis of the marketing potential of new instrumentation: up to now, a large Europe-wide enquiry was performed, aimed at the knowledge of the present situation relative to human motor function

assessment in various clinical environments, and the analysis of real users' needs . Two questionnaires, respectively focused on Rehabilitation Centres and on Hospital Services, were sent to 450 institutions all around Europe. One hundred answers were obtained. They showed real need of quantitative motor functional assessment. At the same time current uncertainties on the equipment policy of the institutions became apparent. The importance of the informative role of CAMARC and the correctness of the chosen research targets were clearly put into evidence.

6. The definition of functional requirements for new devices: a preliminary work has been carried out with respect to devices aimed at the long term monitoring of the Upper Body attitude and at the attainment of temporal and geometric parameters of Gait through an instrumented math of large dimensions and easily installable. The final assesment of the functional requirements for these or equivalent devices is a short term target of the project.

The whole AIM programme is aimed at the enhancement of Medical Bioinformatics in Europe and at the corresponding improvement of Health Care. In such a context the most important perspective of CAMARC is the promotion and the establishment of a European Network of MA laboratories, incorporated both in Research and clinical institutions. They should be able to share data, programs, expertise and Knowledge, on the basis of concerted clinical and experimental protocols. A complementary target is the standardisation of the procedures for testing equipment and the establishment of uniform protocols for data communication.

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BASIC FES

Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

CROSS–TALK BETWEEN FASCICLES USING INTRAFASCICULAR ELECTRODES FOR FES

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INTRODUCTION

For FES it is important to be able to stimulate muscles or parts of a muscle selectively for an accurate control of movement. An interesting way to achieve muscle selectivity might be fascicle selective nerve stimulation.

Human peripheral nerves are composed of a large number of fascicles, kept together by a connective tissue, the epineurium. Number and size of the fascicles vary along the nerves, because the fascicles fuse and divide repeatedly. By this redistribution of nerve fibres among fascicles, the composition of fascicles gets more muscle–specific towards the periphery (1).

It seems obvious to make use of this anatomical subdivision of peripheral nerves for the selective stimulation of muscles. It has been shown by simulations (2) and experiments (3) that intrafascicular wire electrodes are suited for this purpose. The simulations now have been extended: theoretical recruitment curves are calculated for the fascicles. In this paper cross–talk between the fascicles is discussed into more detail. Results of simulations and experiments are compared.

MATERIAL AND METHODS

Simulations

A three–dimensional model of tissue was used (2), composed of wedge shaped volume elements, length of a wedge = 420 μm (z–direction), side of its triangle = 42 μm. Fig.1 shows a cross–section of rat sciatic nerve. The conductivity of the fascicles was assumed to be anisotropic: $\sigma_x = \sigma_y = 0.08\ (\Omega m)^{-1}$ and $\sigma_z = 0.5\ (\Omega m)^{-1}$, the nerve fibres lying in the z–direction. The conductivity of the epineurium was 0.1 $(\Omega m)^{-1}$. The conductivity of the perineurium was $8.4 \cdot 10^{-4}\ (\Omega m)^{-1}$, derived from a paper of Weerasuriya et al (4). The thickness of the perineurium was taken 5% of the

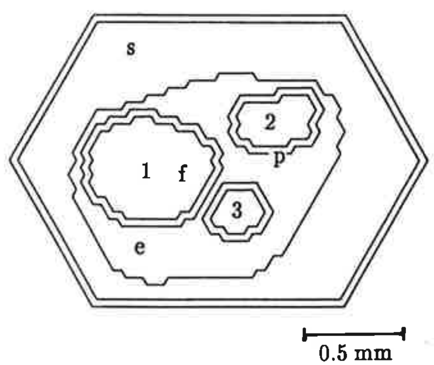

Fig.1. Cross–section of the model (x,y–plane) composed of triangles; f= fascicle, p=perineurium, e=epineurium, s=surrounding medium.

diameter of a fascicle (1). The surrounding medium had a very low conductivity, $10^{-6}\ (\Omega m)^{-1}$, because the nerve was surrounded by air in the acute experiments.

A small cathode was positioned in a fascicle for monopolar stimulation. The potential distribution within the volume–conductor model was calculated by the variational method. The equations were solved numerically by Gauss elimination and Gauss–Seidel iteration.

Nerve fibre excitation was modeled using the network description of a myelinated nerve fibre in an external electrical field, introduced by McNeal (2,5). Cathodic rectangular pulses were used, pulsewidth 60 μs.

Recruitment curves were calculated for a fascicle, which show the fraction of recruited fibres of a certain diameter as a function of stimulus amplitude. The distribution of nerve fibres in the fascicle was assumed to be uniform.

Results are presented by compound recruitment curves. They represent a weighted sum of a number of recruitment curves from several fibre diameters, normalized to maximum force (6). The weight factors depend on the diameter distribution and on the relation between nerve fibre diameter and twitch–force (or tetanus–force) of its motor unit. Weight factors, based on the diameter distribution of α motor–fibres in the EDL of the cat, are shown in table I.

TABLE I

Weight factors for calculation of compound recruitment curves

D(μm)	9–10	10–11	11–12	12–13	13–14	14–15	15–16
Weight factor	0.01	0.03	0.12	0.51	1.00	0.38	0.07

Two parameters were derived from the recruitment curves to illustrate cross–talk:

$$C = \frac{V_m(\text{recr curve})}{V_m(\text{ref curve})}, \qquad (1)$$

where V_m is the stimulus amplitude to attain half of maximum force for a recruitment curve or a reference recruitment curve. C is a measure of the distance between two recruitment curves, not taking into account differences in slope;

$$R = \frac{V_{min}(\text{recr curve})}{V_{max}(\text{ref curve})}, \qquad (2)$$

where V_{min} is the minimum threshold of the nerve fibres for a recruitment curve and V_{max} is the maximum threshold of the nerve fibres for a reference recruitment curve. The two recruitment curves overlap when $R < 1$.

Experiments

The sciatic nerve in the upper hind leg of the rat, consisting of two or three fascicles, was used for stimulation experiments (3). This nerve bifurcates into the tibial nerve and the common peroneal nerve. The common peroneal nerve innervates the extensor digitorum longus (EDL) and tibialis anterior muscle.

The sciatic nerve was stimulated monopolarly by intrafascicular wire electrodes (ϕ 25 μm), isolated except at the tip. Cathodic monophasic rectangular current pulses, having a pulse width of 60 μs, were used. The pulse amplitude was varied during the experiments. Twitch forces of the EDL were measured.

RESULTS

Fig.2 shows a number of compound recruitment curves. Values of some parameters are given in table II. Stimulus thresholds of nerve fibres were low when the cathode was in the same fascicle (two curves at the left). The recruitment curve of the large fascicle 1 lay at higher stimulus amplitude than the curve of fascicle 2.

The other recruitment curves are curves for neighbouring fascicles of the one containing the cathode. When the cathode was in fascicle 1 (solid lines), the lowest threshold in fascicle 2 was about 20 % higher than the highest threshold in fascicle 1. So the curves did not overlap.

However when the cathode was in the (small) fascicle 2 (dashed lines) the distance between the curves was larger: the lowest thresholds in fascicle 1 and fascicle 3 were 2.1 respectively 1.4 times higher than the highest threshold in fascicle 2.

Also shown in fig.2 (dotted curve) is the effect of modeling a layer of saline, $\sigma = 2.0\ (\Omega m)^{-1}$, on the outside of the nerve. The cathode being in fascicle 1 the stimulus thresholds of the fibres in fascicle 2 became a factor 1.7 higher.

Experimental results are shown in fig.3. When the cathode was placed in the fascicle passing into the common peroneal nerve rather different recruitment curves were obtained, C=2.3 for the curves marked by □ and ○ . C=7.7 for the curves □ and × .

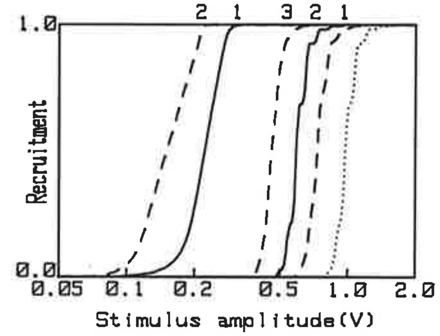

Fig.2. Compound recruitment curves, fascicle number above, cathode in centre of fascicle 1 (———) or 2 (– – – –). For the dotted curve for fascicle 2 a layer of saline was modeled outside part of the nerve, cathode in centre of fascicle 1.

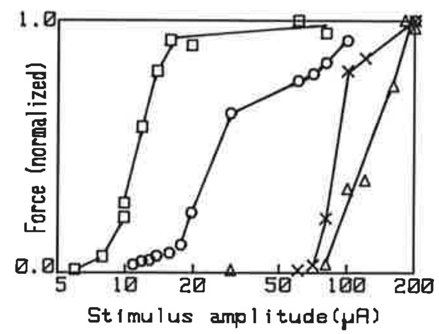

Fig.3. Experimental recruitment curves of the EDL. The cathode was positioned in the fascicle of the sciatic nerve passing into the common peroneal nerve (□ , ○) or in another fascicle (× , △).

TABLE II

Parameters C and R from compound recruitment curves. First two columns describe one curve. The reference curve is calculated for a fascicle containing the cathode.

Recr curve of fascicle	Cathode in fasc	Reference recr curve	C	R
2	1	1	2.6	1.2
3	1	1	2.3	1.1
1	2	2	4.8	2.1
3	2	2	3.0	1.4
2(s)	1	1	4.4	2.1
2	1	2	3.8	1.9
2(s)	1	2	6.3	3.1

DISCUSSION

It can be concluded from simulations and experiments that it is possible to stimulate fascicles selectively by intrafascicular wire electrodes.

Recruitment curves and also cross–talk depended on the dimensions of the fascicles. For the fascicle containing the cathode the recruitment curve shifted to higher stimulus amplitude when the dimensions of the fascicle were increased. For neighbouring fascicles the recruitment curves shift to lower stimulus amplitude when the dimensions of the fascicles were decreased, because the thickness of the perineurium also decreased. So overlap of recruitment curves may arise when the cathode is in a large fascicle with small fascicles in its neighbourhoud.

Experimentally rather different recruitment curves were obtained when the cathode was in the fascicle entering the common peroneal nerve (C=2.3). From simulations (not shown) it appeared that this can only partly be due by variations of the position of the cathode in the fascicle. Another reason for variations might be a varying amount of saline along the wire electrode giving a leakage current from cathode to outside of the fascicle.

Cross–talk was different between simulation (C=3.8) and experiment (C=7.7). This might be caused by geometrical differences or by the choice of parameters in the simulations or by presence of saline on the outside of the nerve in the experiments. Simulations showed that C increases when the conductivity of the surrounding medium is increased.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

FREQUENCY RESPONSE MODELS OF NINE MUSCLES IN THE CAT'S SHANK

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INTRODUCTION

The design of a Functional Electrical Stimulation (FES) based high performance joint control system requires controllers and feedback devices which compensate for the dynamic behavior of the muscles involved in the movement. In order to do so, dynamic models describing the dynamic response of active muscles are necessary. Agonist, antagonist and synergistic muscle groups can be expected to move the joint under control while keeping its mechanical integrity and stabilizing the proximal and distal joints. It is necessary, therefore, to explore the architectural and functional characteristics of a muscle in relation with their effect on its dynamic performance.

The viscoelastic stiffness of a muscle may change according to the tendon length, the pennation type, muscle length and weight and result in changes of the dynamic response. In addition, the fiber composition may influence the speed of response of the muscle, since it is widely accepted to affect the course of single twitches.

A previous report (Baratta et al., 1990) concluded that control strategies which included the orderly recruitment of motor units as part of their activation paradigm had little influence on the dynamic behavior of muscle.

The objective of this study is to determine the dynamic response model of several muscles in the cat's hindlimb. Those muscles exhibit a variety of architectural properties, and it is expected that those properties which play a role in determining a muscle's response may be identified.

METHODS AND RESULTS

Fifteen adult cats were anestetized with chloralose (60 mg/Kg). Their sciatic nerve was exposed and all branches denervated except those to the muscle under study. Two bipolar cuff electrodes were placed on the nerve for later connection to the stimulation system. Pins were placed through the femoral condyles and mid tibial shaft for rigid fixation. The muscles' tendinous insertion was freed, and attached to a GRASS FT-10 transducer upon mounting the animal on a rigid platform. The ankle joint was disarticulated to allow direct line of connection of anterior muscles.

The stimulation system has been described and validated in detail (Zhou et al., 1987; Baratta et al., 1990a,b). Briefly, an IBM-XT computer generates two voltages, one of which is used as input to a Voltage Controlled Oscillator (VCO), in charge of controlling the firing rate of the active motor units. The second voltage is used by a pulse modulator with a rate of 600 pps, to induce the orderly recruitment and derecruitment of motor units. Concurrent, but, independent control of motor unit recruitment and firing rate were achieved. These conditions approximate voluntary contractions more closely than is otherwise possible. Sinusoidal motor unit recruitment coupled with increase in the firing rate produced sinusoidal variations in the muscle force. Stimulus levels were calibrated such that at 0.4 Hz, the force would range from 20 to 80% of the maximal available force. Trials were repeated at .6, .8, 1, 1.2, 1.4, 1.6, 1.8, 2, 2.5, 3, 4, 5, and 6 Hz. Force and stimulus voltage were sampled at 64 Hz, windowed, and put through an FFT algorithm. A point in the frequency response was obtained by the follow equation:

$$\frac{D}{V} = \frac{D}{V}/\emptyset$$

where D and V are the FFT of the force and stimulus input voltage at the trial's fundamental frequency and Ø is their phase difference.

Gains were normalized with respect to the .4 Hz gain, converted to dB, and the data plotted in conventional Bode gain and phase plots. The method of least squares was used to obtain best fit transfer functions which were a linear second order critically damped model with a time delay. The results of the best fit for each muscle are shown below:

Muscle	Double pole Location (Hz)	Td (MS)	Muscle	Double pole Location (Hz)	Td (MS)
L. Gastroc.	1.55	12	FLDig.L.	2.15	10
Soleus	1.8	16	Tib.Post	2.15	12
M.Gastroc.	2.0	8	Ex.Dig.L.	2.5	9
Peron. Brev.	2.1	9	Tib.Ant.	2.8	17
Peron.Long	2.1	8			

Multivariate stepwise regression analysis revealed that the functional location of the muscle in the shank with respect to its role as flexion/extension or inverter/everter were the

most important factors in determining the location of poles for the model. Also of importance was the pennation type. Using a coding system where posterior and medial location were indexed (1 for most posterior/medial, 9 for least posterior/medial) and where architecture is coded (1 for fusiform, 2 unipennate, 3 bipennate), a predictive equation for pole location in the cat's shank can be obtained:

$$P = 1.747 + .152 * PI - 0.047 * MI - 0.094 * A$$

$$R = .091, SE = .0162$$

DISCUSSION

The most important fact emerging from this data is that skeletal muscles with widely varying architectural properties have dynamic response models which are vastly different in pole location (1.55 Hz double pole in the L.G. vs 2.8 in the TA). The second order critically damped response allows miminum rise time without overshoot, this optimizing the response speed.

Size related properties such as mass, muscle length, and tendon length had no bearing on the pole location, whereas functional properties such as location with respect to flexion/extension (posterior index) or inversion/eversion (medial index) and pennation type did correlate with pole location. It was surprising that fiber composition was not correlated with dynamic response, since this factor generally determines the muscle's twitch response. The difference may be due to the orderly and progressive activation/deactivation of motor units, which is not present in twitch responses or in previous stimulation systems.

In summary, the data indicates that the functional responsibility and pennation of a muscle contribute to its dynamic response significantly, whereas size and fiber-related parameters do not. A pole predictive equation may be obtained by combining the location and functional parameters with the muscle's architecture. These data may help in the design of successful neuro-prosthetic motor control systems.

ACKNOWLEDGMENT: This work is supported by NSF Grants EET-8613807 and EET-8820772.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

PHYSIOLOGICAL COST INDEX OF PARAPLEGIC LOCOMOTION USING THE ORLAU PARAWALKER

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INTRODUCTION

Energy expenditure is an important parameter in the assessment of orthotic treatment of heavily handicapped patients such as paraplegics. The most commonly used physiological parameter for assessment of energy expenditure has been the measurement of Oxygen Uptake. This technique involves either wearing a noseclip and breathing through a mouthpiece or breathing through a snugly fitting mask so that expired air can be collected into a Douglas bag for analysis. It can also be directly analysed with the Oxygen Consumption Meter(e.g. Oxylog). However, in clinical situations involving paraplegics this method proves to be impractical; also, not all the institutions involved in walking rehabilitation of paraplegics have respiratory laboratory facilities.

In able bodied subjects heart rate and Oxygen uptake have a linear relation up to submaximal workloads(4). This has enabled clinicians and rehabilitation engineers to monitor the energy cost of a variety of physical activities by monitoring heart rate alone. Sympathetic regulation of the cardio-vascular system has been claimed by various workers to be between thoracic spinal cord levels 1 and 6(5,6,7,8,9). This implied that monitoring heart rate as an indicator of energy expenditure was unreliable in cases of upper thoracic level paraplegics.

A study involving forty four traumatic complete paraplegics with lesion levels ranging from Th3 to Th10 was carried out by Bar-On & Nene(1). Subjects were put through an arm cranking exercise routine with increasing power levels. Their heart rate and oxygen uptake was measured for each power level. The findings of that study suggested that the sympathetic contribution to the cardiac plexus is intact in traumatic complete paraplegics with a lesion level below Th3.

In the light of this knowledge the method of measurement of heart rate can also be extended to walking rehabilitation of paraplegics. It has been shown in the past that it was possible to establish walking performance of patients by monitoring speed and heart rate(10,11). MacGregor(2,3) described a method of combining the two parameters to produce a single index called Physiological Cost Index (PCI) as an indicator of locomotor efficiency.

One of the main interests of ORLAU has been using orthoses for Paraplegic Locomotion. The task of comparing the energy cost of locomotion using the various forms of orthoses and of monitoring the progress of training can be made much simpler by monitoring heart rate alone.

The purpose of this study was to establish a baseline range of PCI for thoracic level traumatic complete paraplegics ambulating with the ORLAU ParaWalker.

SUBJECTS

Sixteen subjects participated in this study. There were 14 males and 2 females. All were post trauma complete paraplegics with lesion levels varying from Th3 to Th12. Their age ranged from 25 years to 50 years. Duration of paraplegia varied from 3 years and 7 months to 21 years and 9 months. All were experienced ParaWalker users. Duration of their ParaWalker use varied from 6 months to 6 years and 3 months. Eight subjects used their ParaWalker once a week. Three subjects used it 3 times a week. Two subjects used it 4 times a week. One subject used it 5 times a week and the remaining two subjects used it every day of the week. Their weight plus ParaWalker ranged from 50.9 kg to 85.9 kg. None of them had any symptoms related to their cardio-pulmonary function either in the past or at the time of the test.

METHOD

On arrival in the department subjects were asked to don their ParaWalker. They were weighed in their orthosis. Electrodes were applied to their chest wall. Wires were attached to the electrodes. A backpack containing a signal amplifier, digital encoder and Infra-Red signal transmitter was strapped on and the wires from the electrodes were attached to the backpack.

The subject then rested in his wheelchair. Heart rate was monitored for 10 minutes to establish the resting heart rate. Then the subject stood up and waited for a while until the heart rate was stabilised again. On instruction the subject walked along a

walkway, in between timer gates placed 6.1 m apart, at a steady speed. With the help of the computer the ambulatory heart rate and the time taken to traverse the distance between the timer gates was recorded. The subject walked 5 times along the walkway with a 1 minute rest period between each walk. At the end of the 5th walk the subject sat down in the wheelchair and rested. Heart rate was monitored during this recovery period until it lowered to normal resting rate or for a maximum of 10 minutes.

From the resting heart rate, the heart rate at the end of each walk and the time taken to traverse the distance between the timer gates it was possible to calculate the PCI for each walk. The mean of the 5 walks was taken as the PCI of ParaWalker ambulation for each subject. The following formula was used to calculate the PCI:

$$\text{PCI} = \frac{\underset{\text{(beats/min)}}{\text{HR at the end of a walk}} - \underset{\text{(beats/min)}}{\text{Resting HR}}}{\text{Speed (m/min)}}$$

beats/m)

RESULTS

Walking speed of the subjects ranged from 0.168 m/s to 0.442 m/s. The mean speed was 0.283 m/s. The mean Physiological Cost Index was 3.121 bts/m. It ranged from 1.464 bts/m to 4.752 beats/m.

DISCUSSION

The Physiological Cost Index in the present study group showed wide variation. It did not show any correlation to the level of spinal cord lesion. There was also no correlation to the weight of the subject. It also did not show any correlation to either the duration of ParaWalker use or to the frequency of usage. Nene & Patrick(12) measured the oxygen uptake of ParaWalker users to calculate the energy cost of their locomotion. In that study as well there was no relationship found between the level of lesion, duration of ParaWalker use and energy requirements.

The difference in resting and walking heart rates will depend upon the physiological status of an individual and the walking speed will depend upon the efficiency of the orthosis in use. It will be an indicator of energy expenditure for an individual using a different orthosis or trying out modifications to the same orthosis. It will also indicate his physical status using the same orthosis during the training period. Direct comparison of the PCI of an individual using the ParaWalker to the PCI of an another individual using a different orthosis is unrealistic. But it would be statistically reasonable to compare the range of PCIs of a group of subjects using the ParaWalker to PCIs of an another group of subjects using different orthoses e.g. bilateral long leg braces or Reciprocating Gait Orthosis.

The findings of this study establish a base line range of PCI for the ParaWalker group. This study enables the researchers working with the Functional Electrical Stimulation (FES) to compare the effect of various strategies of hybridisation of the ParaWalker. As mentioned before, researchers working with the other forms of orthosis can compare the efficiency of their orthosis to the ParaWalker.

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CLINICAL FES

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

RECENT ADVANCES IN CLINICAL FES

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INTRODUCTION

A great advance in computer technology enabled to fabricate a portable size FES system with multichannel outputs. This has made possible to apply the FES system clinically for restoring motor function of the paralyzed upper and lower extremities in the spinal cord injury (SCI) and stroke patients. Strictly saying, however, controllability of the FES system is not satisfactory yet at present and, therefore, FES for the extremities has not been accepted widely. Nevertheless, FES has proved to provide significant improvement in motor function of the paralyzed extremities. Great efforts have been made toward attaining wide-range clinical usage of the FES system by many investigators.

In this paper, I will discuss about recent advances of clinical FES for the paralyzed extremities and objectives and problems in the coming year.

ELECTRODE

The stimulus output is applied to the muscle branches of the peripheral nerves through electrodes. Clinically available electrodes at present are surface, intramuscular, epimysial and epineural electrodes. Surface electrodes have been used widely, especially in the field of lower extremity FES, because of easy installation without surgical operation. Since finer control of joint motion such as FES for the hand requires selective activation of individual muscles, implanting electrodes are advantageous in this respect.

Most of the clinical application of FES for the hand has been achieved by using percutaneously indwelling intramuscular electrodes [1,2]. The electrode design is primarily a helical coil wound from a Teflon-coated multistrand 316L stainless steel rope that is deinsulated at the tip for stimulation (3). A great problem preventing wide-range clinical usage of the percutaneous intramuscular electrode is electrode failure due to metal fatigue. Peckham et al reported the rate of failure in upper extremity FES was about 5% per month [4,5]. In the lower extremity FES for the paraplegic patient, 35% of the percutaneous intramuscular electrode were broken within the first four months [6]. In order to increase fatigue resistance of the electrode, we developed the percutaneous intramuscular electrode with high mechanical strength, elasticity and flexibility [3]. Although the design of this electrode is the same of the above mentioned percutaneous electrode, A 19 strand rope wound from 316L hard drawn stainless steel wires (diameter of each wire is 25 microns). The rate of fracture was below 2% per year even in the lower extremity FES. Thus, percutaneous intramuscular electrode was clinically available.

Epimysial electrode is a disc type electrode sutured the epimysium of the muscle used by a research group of Case Western Reserve University (CWRU) [5]. This electrode is connected to a totally implantable FES system and, therefore, the rate of electrode failure and migration is very low. In addition, such implantable system is advantageous for elimination of maintenance of the percutaneous site.

A group of University of Vienna has clinically applied a total implant system [7]. In this system, four electrodes were attached to the epineurium of the nerve branches for the control of the locomotive movement of the paraplegic.

FES FOR THE UPPER EXTREMITY

Clinical FES for the upper extremity has been potentially performed on quadriplegic patients caused by cervical cord injury (CCI). In CCI patients, extent of their paralysis of the upper extremity depends upon the level of injury. Volitional finger movements are mainly impaired in C6-7 quadriplegia. In C5 quadriplegia, the patient with a injury at C5 loses voluntary control below the elbow. Almost all of the muscles of the upper extremities are paralyzed in C4 quadriplegia.

Peckham et al. in CWRU have applied a percutaneous FES system and an implantable FES system with epimysial electrodes which provide up to eight channel outputs in order to restore the hand function in the C5 and C6 quadriplegic patients [1,4,5]. They controlled two grasp patterns of the hand, lateral prehension (or key grip) and palmar prehension (or three jaw chuck pinch). Muscles for stimulation were selected for decreasing the number of the electrodes. They did not implanted the electrode to the intrinsic hand muscles because of difficulties of implantation and control. In addition to FES, some surgical procedures such as arthrodesis, sutures of the tendons (synchronization) of the finger flexors and/or extensors and tendon transfers were performed for getting excellent controllability and reliability of the joint movements [8]. Orthoses were also used for stabilizing the wrist in C5 quadriplegia. The stimulation patterns were determined empirically.

Well coordinated movements of the wrist and hand for grasp and release in C5 quadriplegia were provided by a percutaneous portable FES system which stored stimulation data created from EMGs of normal volunteers during motion [2,9]. Linked movements of a normal hand and wrist system could be simulated by applying the programmed stimulus simultaneously to the hand and wrist muscles in the C5 quadriplegic patients. In addition, we also implanted the percutaneous intramuscular electrodes to the interosseus dorsales (DI) and palmares (PI) [10]. Simultaneous stimulation to PI and DI induced extension of the proximal and distal interpharangeal joints (PIP and DIP joints, respectively) of the fingers. The intrinsic stimulation provided well-balanced functional positioning of the metacarpopharangeal(MP), PIP and DIP joints adjusting to the size and shape of the grasping objects. Thus, good hand function for activities of daily living (ADL) could be achieved with a 16 channel portable FES system.

FES control of the totally paralyzed upper extremity in C4 quadriplegics have been achieved by surface and percutaneous FES systems. Nathan examined motor points of muscles in the upper extremity and made maps for attaching the electrode to the skin [11]. He controlled all of the joints except the shoulder in the upper extremity by a voice-controlled surface FES system with the aid of a suspension for the arm [12]. We implanted the percutaneous intramuscular electrode to almost all of the upper limb muscles below the glenohumeral joint [2]. Under the usage of a balanced forearm orthosis for supporting the arm against gravity and allowing volitional movement of the upper arm, the C4 quadriplegic patients could control their upper extremities for ADL by using their respiratory control commands.

FES FOR THE LOWER EXTREMITY

FES-controlled standing and walking in SCI patients have been also performed by surface and implantable systems.

Kralj et al. in Yugoslavia have clinically experienced the restoration of locomotive function in the paraplegics by using a surface FES system more than ten years [13]. A four channel FES system was used and four sets of surface electrodes were attached bilaterally to the quadriceps femoris and the common peroneal nerves. He showed erect standing obtained by the quadriceps femoris stimulation under the usage of a wheel chair attached folding frame for balance. In FES-induced locomotion, stance and swing phases were alternately achieved by stimulation of the quadriceps femoris and common peroneal nerve, which induced a withdrawal reflex of the stimulated lower limb, respectively. The paraplegic patient could walk at velocities from 0.2 m/s up to 0.3 m/s on distances of 100-200m with a help of a walker or crutches.

A number of a hybrid FES system which comprises a brace and an FES system has been reported. Recently, Solomonow in the United States used a reciprocating gait orthosis (RGO) [14] and Andrews in UK developed a mechanically passive supracondylar knee ankle foot (SKAFO) brace [15] in combination with the surface FES system in the paraplegics. Ichie et al. in Japan also developed a percutaneous hybrid FES system which comprised long leg braces (LLB) and a percutaneous FES system [16]. In the hybrid FES, activation of the extensors of the lower limbs can circumvent and, therefore, reduce muscle fatigue and energy consumption during gait control. The hybrid FES provided locomotion for up to 4.25 hours at velocities of 0.15 m/s to 0.35 m/s.

A percutaneous FES system has also applied to the locomotion control of the paraplegics [17,18]. Marsolais et al. in CWRU implanted percutaneous intramuscular electrodes to the muscles of the trunk, hip, knee and ankle [18]. They demonstrated forward and backward locomotion, stair ascending and descending and other locomotive movements by 42 channels FES system. They reported that 800 meters of maximum walking distance and 1.0 m/s of maximum speed were achieved by this percutaneous FES.

Vienna group developed totally implantable FES system and applied clinically to four paraplegic patients for 6 years [7]. Round-about electrostimulation by attaching four electrodes to the epineurium of one peripheral nerve in a square position was adopted for activating paralyzed muscles. With the aid of crutches, two paraplegics could walk with a 4-point gait for about 100 meters and the others walked with a swinging through gait.

DISCUSSION

Selective and reliable activation of the paralyzed muscles by implantable electrodes enables to restore motor function of the upper and lower extremities in the upper motor neuron disorders. In addition, development of totally implantable FES systems would be a final goal for clinical FES. However, it seems dangerous to apply the total implant system to the patient in a hurry before establishing control strategies of FES. Since FES is a powerful means for restoration of motor function of the paralyzed extremities, accumulation of basic research works and clinical experience would be necessary.

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Electromyographical Kinesiology
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TARGET AND OVERFLOW ACTIVITY OF THE UPPER LIMB MUSCLES DURING STIMULATION BY SURFACE ELECTRODES

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INTRODUCTION

FNS for restoration of hand or upper limb function has been carried out in a number of research centres, both using implanted electrodes [1-4], and surface electrodes [4-8]. The stimulation can activate not only the target muscle , but also unwanted overflow muscles in the limb. This is particularly a problem when using surface electrodes. For fine control of hand prehension and release, and for control of positioning and orientation of the hand, maximum activation of the target muscle is required with minimum overflow. Positioning of the electrodes and their geometry are the principal factors which control the ratio of target muscle to overflow muscle activation level.

In this work the activation of target and overflow muscles are reported for close to optimum electrode conditions. Maximum resolution of muscle activation is obtained from minimum electrode size. Limitations to this size depend on the type of patient and stimulation system. The research reported here was carried out on, and is intended for C4 quadriplegic subjects only. Here sensory feedback from the forearm and hand surface is absent, and pain is not a problem when applying high currrent densities through the electrodes.The current density is limited instead by burning of the skin. For the stimulation parameters used, danger of burning has been found to occur at current densities greater than 50mA/cm^2 [7]. Minimum electrode size for the maximum current intensity is used. Methodical techniques are used to position the electrodes for each muscle to elicit maximum ratio of target to overflow muscle activity [9].

Current-controlled stimulation gives a more stable level of muscle activation due to its independence from interelectrode impedance. Current-controlled stimulation is used throughout this research.

APPARATUS & METHODOLOGY

Monophasic compensated double square-wave pulses having 0.3mS width, and 0.7mS delay at 30Hz frequency are current-controlled both manually, and through a microcomputer DAC. The stimulation is delivered through a bipolar search probe, each leg of which is tipped with a conductive rubber electrode, 6mm x 14mm arranged colinearly. Electrical contact with the skin is made with commercial gel.

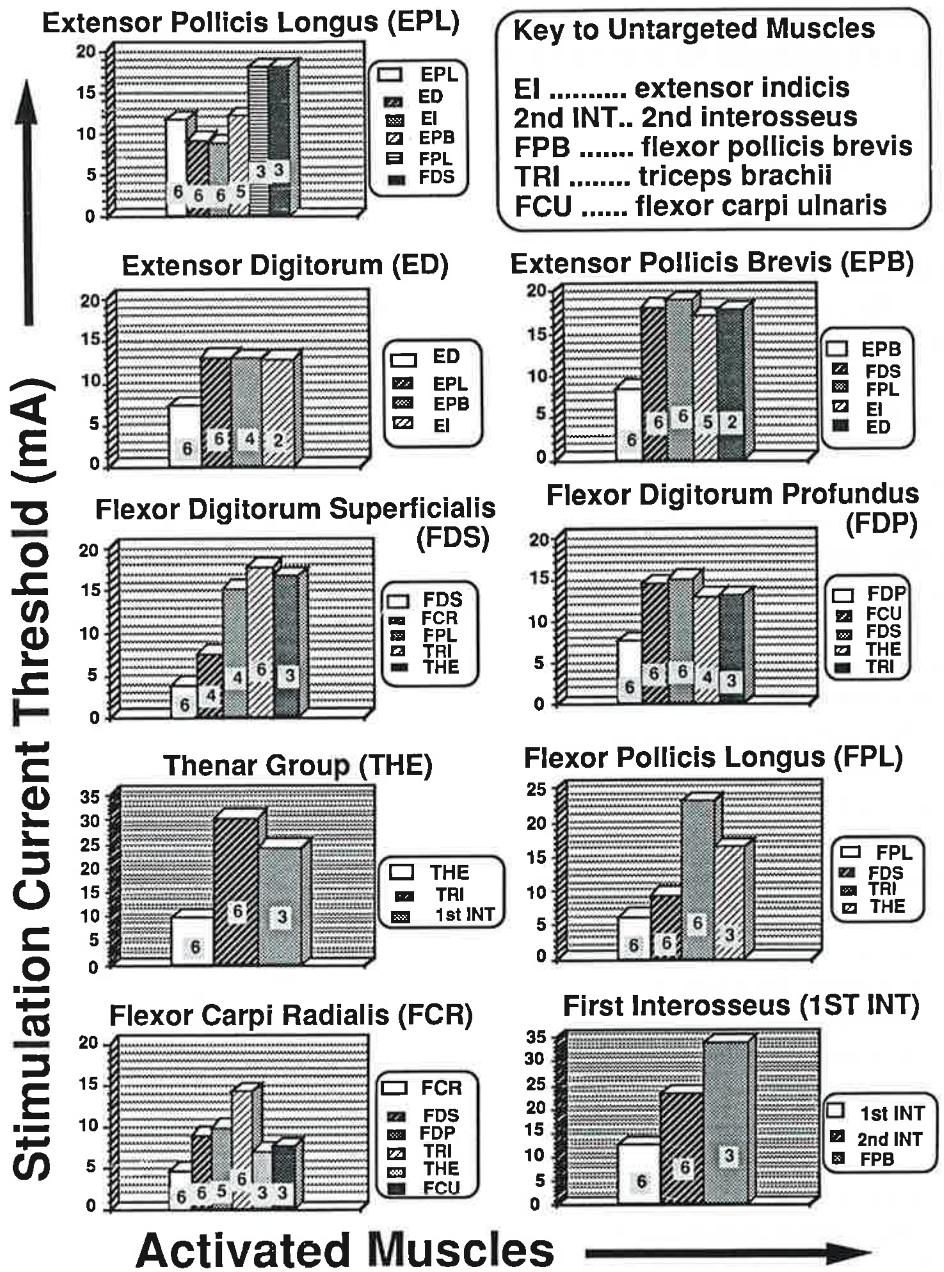

Figure 1 Current Thresholds for Target and Overflow Muscles

Joint moments are monitored isometrically by force sensors (Interlink Electronics, California) attached to a glove, constrained by moulded polymer shells. The results are monitored on the microcomputer through the A/D. Threshold activation of the muscles is monitored by manual palpation of the muscle tendon, by visual observation of the muscle body and the motion of the distal segments.

The single test subject is a C4 quadriplegic, with some atrophy of the upper limb musculature. The subject has been receiving stimulation once per week for four years. The limb was suspended during testing in a mid-range position with the forearm and hand horizontal, palm downwards. The muscles were tested unfatigued.

RESULTS

Figure 1 shows the stimulation current threshold for targeting seven forearm muscles, and two hand intrinsic muscles. The current thresholds of the overflow muscles are shown to the right of each target muscle. The measurements were repeated on six seperate occasions. The graphs show the averaged results, with the number of times the overflow muscle was activated marked for each muscle.

In figure 2 the maximum moment generated in the principal distal joints is shown for isolated activation of the same target muscles. Also shown is the moment range in which the activation is contaminated by acceptable overflow, not to impede hand prehension/release or limb motion (e.g. overflow to a finger flexor muscle when targeting the thumb flexor). For some muscles the largest generated moment shown is contaminated by unacceptable overflow (e.g. to the Triceps Brachii).

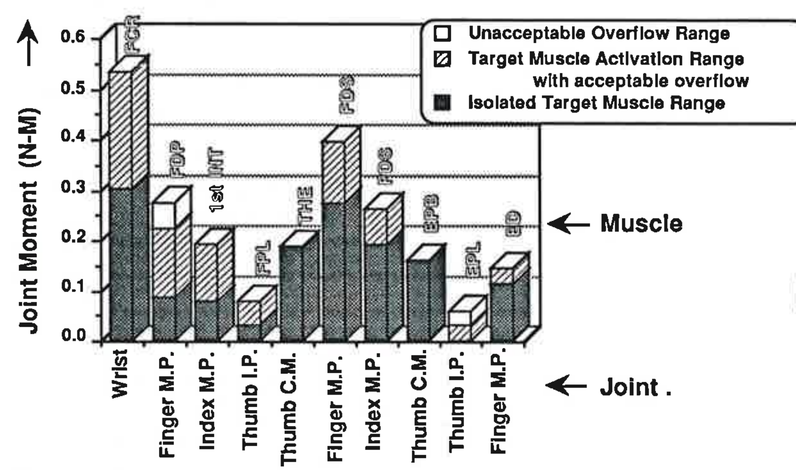

Figure 2 Joint Moment Ranges

DISCUSSION

The measurements have been carried out on one subject. The results would tend to vary between individuals, depending on the physiological condition of the limb and the accuracy of electrode placement. Atrophy tends to improve access to the deep-lying musculature, but to generally decrease the uncontaminated activation range of each target muscle, and also the level of joint moment generated. For a given subject and electrode placement, fatigue also introduces time-dependecy into the results. As the target muscle fatigues its current intensity threshold tends to rise. Overflow muscles usually remain relatively fresh, and as a result the useful working range of the muscle's moment generating capability tends to decrease, often to zero. This has been observed in practice where complex computer-controlled stimulation-generated functional movements become too contaminated with overflow when the limb is fatigued. This limits the duration of a working session. FNS training of the muscles increases their fatigue resistance; we have found the untrained limb to be capable of generating useful arm function for approximately 15 minutes, and the trained limb for several hours.

ACKNOWLEDGEMENTS

The author wishes to thank Operation Wheelchairs Committee of London for funding the equipment for this research, the SCI Rehabilitation Centre, Tel Hashomer, and to Batia without whom this research would not have been possible.

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P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

NEUROMUSCULAR STIMULATION IN C4-QUADRIPLEGIA: CLOSED LOOP CONTROL OF THE HAND FOR WRITING

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INTRODUCTION

Functional Neuromuscular Stimulation (FNS) has in the last few years been successfully applied to the upper limb in several centres, both for generation of hand prehension/release in C5 and C6 quadriplegia [1], and for closed-loop control of gripping force [2-4].

Computer generated movements of the whole upper limb in C4 quadriplegia has been carried out in two centres [5-9]. Stimulation generated prehension/release patterns, vertical hand movements from wrist flexion/extension and outward reach from elbow extension are triggered by voice commands in the Beer Sheva system [6-9]. The forearm is constrained in an arm support to move in a horizontal plane. Voluntary shoulder girdle movements generated by upper trunk muscles pass along the suspended arm causing voluntary control of the movement of the hand in the horizontal plane. Several activities of daily living have been restored to the quadriplegic subject [7,9]. The system has been to date open-loop, system parameters have been adjusted on-line by voice commands from the subject himself according to visual feedback from his upper limb, its response to the stimulation, and its interaction with the utensils and environment.

One such restored activity has been writing, including picking up the pen from a holder and returning it after use. With the open-loop system this activity is not easy. The quadriplegic subject has been required to control not only the form of the letters, but also the height of the pen above the paper or the pen/paper pressure, and even the gripping force. This left little opportunity for him to enjoy writing, and has effectively limited the length of writing to a few letters at a time.

An enhancement to the system has been developed and tested for closed-loop control of the above parameters. The strength of the gripping force, the raising and lowering of the pen, and the pen/paper pressure are transferred to automatic control, leaving the system user in control only of the pen strokes plus a simple **UP** or **DOWN** command to raise or lower the pen.

METHODOLOGY

A key grip is used to grip a pen. Stimulation of the flexor digitorum superficialis muscle along its medial border generates finger flexion, particularly in the index finger. The extensor pollicis brevis muscle is activated to extend the thumb, opening the hand when the pen is to be taken. The flexor pollicis longus muscle and the thenar group are activated to flex the thumb and close the hand, gripping the pen against the side of the flexed index finger The pen is maneuvered horizontally to the required position on the paper by voluntary shoulder girdle movements. A command **DOWN** triggers stimulation of the flexor carpi radialis muscle which generates wrist flexion causing the pen to lower. A command **UP** reduces the stimulation current intensity to this muscle, and the mechanical spring return causes the wrist to extend and the pen to rise. Where the wrist extensor muscles espond sufficiently, these are activated instead of the spring return. In the open-loop system continual on-line adjustment of gripping force and pen height is required.

The Instrumented Pen Sleeve (figure 1)

The gripping force and pen/paper pressure are monitored during writing. An instrumented sleeve was designed into which a pen is fitted. This allows the pen to be changed when required. Pressure sensitive pads (Interlink Electronics Inc., California) are used as the force monitoring elements. These monitor the gripping force, and the axial and lateral components of the pen/paper force, and are input to the microcomputer through an A/D.

The Closed-Loop Control Software

Control of Gripping A minimum force required to hold the pen during writing is specified, plus a target band width within which the gripping force is allowed to drift uncontrolled. If the gripping force strays outside the target band, the stimulation current intensity to the muscles involved is increased or reduced by constant increments. Two artificial intelligence algorithms are added to correct drift and to "abort" when combinations of events indicate the pen has slipped out of the hand.

Control of Pen/Paper Force A low pen/paper target force is predefined. During writing the measured force is subjected to a fast, high amplitude arbitrary imposed error input due to the friction component of the force in the plane of the paper, generated in the direction opposite the pen stroke. A PID control regime was applied of the form:

$$I = K_1\varepsilon + K_2\frac{d\varepsilon}{dt} + k_3\int\varepsilon dt$$

where ε is the error in the resultant contact force, and k_1, k_2, and k_3 are constants . The system output I is the stimulation current intensity applied to the flexor carpi radialis muscle.

A second and separate operating mode occurs when the pen is in the initially raised position. A constant fast rate of increase of the stimulation current intensity to the flexor carpi radialis muscle lowers the pen quickly. At the moment of pen / paper contact the stimulation current intensity is reduced by a backoff step ΔI_1. This is necessary to prevent overshoot. Thereafter the P.I.D. control regime is applied.

Between pen strokes the pen is raised a few millimetres above the paper and lowered. On input of the command **UP** the stimulation current intensity when the pen/paper force was last close to the target level, is decremented by a step ΔI_2, enough to raise the pen from the paper. The command **DOWN** returns a step increase $k_4\Delta I_2$, where k_4 is slightly less than 1. The sensor drift is zeroed during every command **DOWN**. All the control regime parameters were optimized during experimentation.

Control Heirarchy
1. Voice command input.
2. Gripping force error correction.
3. Pen / paper force control.

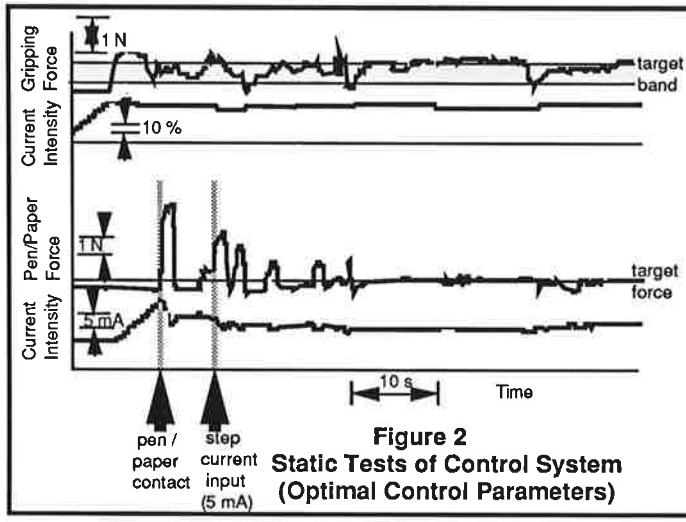

Figure 2
Static Tests of Control System (Optimal Control Parameters)

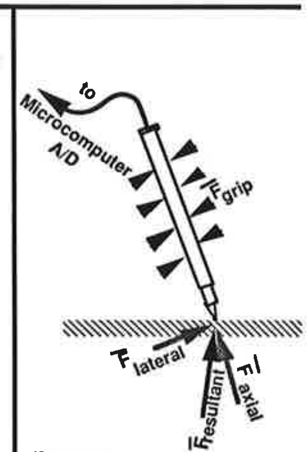

Figure1
The Instrumented Pen

EXPERIMENTATION AND RESULTS

Tests were carried out on one C4-quadriplegic subject. Figure 2 shows the results of a static test: with no voluntary contribution from the test subject, the stimulation activates vertical motion only. A set of parameters close to optimal is shown. The upper graphs show the gripping force and the amplification constant applied to the stimulation current intensity of the muscles involved in the gripping, plotted against time. The lower graphs show the pen/paper pressure and the stimulation current intensity applied to the flexor carpi radialis muscle. After approximately 16 seconds a 5mA current increment is imposed on the system. The response to this can be seen subsequently.

Control system parameters optimized in the static tests were modified for dynamic tests and free writing where a random force input is imposed on the system due to pen/paper friction. The quality and quantity of the free writing improved as the control parameters approached optimum. Several times full pages were written by the quadriplegic subject.

DISCUSSION AND CONCLUSIONS

The writing activity has been substantially improved by the introduction of the closed-loop system. Free writing ability has increased from individual words using the open-loop system, to full pages of writing using the closed-loop system in each sitting. In further work an adaptive control regime is being assessed to increase the flexibility of the system.

ACKNOWLEDGEMENTS

The author wishes to thank Operation Wheelchairs Committee of London for funding the equipment for this research, the Spinal Cord Injury Rehabilitation Centre, Tel Hashomer for providing facilities, and most of all to Batia without whom this research would not have been possible.

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MUSCLE FATIGUE

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

ASSESSMENT OF DYNAMIC MUSCLE FATIGUE BY EMG- AND KINEMATIC PROFILES.

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INTRODUCTION

Human muscle fatigue has predominantly been studied under controlled isometric and/or isokinetic conditions. The electromyogram (EMG) has been used as a non-invasive technique for the assessment of muscle fatigue during these conditions. In everyday life, muscle fatigue is not a problem related to isometric/isokinetic conditions but related to e.g. sport and occupational work, and as such to dynamic muscle work. Very few studies have, however, tried to quantify dynamic muscle fatigue by EMG recordings. It is well-known that many injuries occur when the muscles are fatigued, but the extent to which muscle coordination or muscle performance fails has not been studied. The attempt in the present study is to quantify dynamic muscle fatigue by changes in both muscle coordination and muscle performance.

MATERIAL AND METHODS

Ten healthy soccer players (10 males, mean age 25+/-1 years) participated. All volunteers walked (5 km/h) until endurance on a treadmill with a 25% incline. Surface EMGs were recorded from the right hamstring lateralis (m. biceps femoris), hamstring medialis (m. semitendinosus), m. vastus lateralis, m. gastrocnemius medialis, m. soleus, and m. tibialis anterior. The EMG was pre-amplified ten times by small headstages (model 2010, Intronix Technologies, Ontario, Canada), transmitted to the amplifiers, and filtered (20-450 Hz) before data acquisition. A pressure-sensitive transducer was mounted externally under the heel. The positions of the upper and lower legs were monitored by attached inelastic wires connected to a constant-moment spiral spring. The velocity was calculated numerically. The EMG onset and offset times of the bursts were determined with respect to heel contact, and the burst duration was calculated. For each EMG, burst the root mean square value (RMS) and the mean power frequency (MPF) were calculated. The power spectra of the individual EMG bursts were estimated by a 10'th order autoregressive model. The EMG parameters were calculated as the mean of the 30 first and 30 last steps. The EMG was full wave rectified and smoothed before profile calculation. The EMG and kinematic profiles were normalized to 100% stride time, and the mean amplituede was normalized to a 100% (1). Students t-test was used for statistical analysis.

RESULTS

The mean time to endurance was 3.1+/-1.5 min (+/-SD). The peak velocities and the peak positions of the upper and lower leg were changed marginally (3-9%) and the patterns of the kinematic profiles remained constant.

TABLE I.
EMG CHANGES DURING FAST UPHILL-WALKING

The mean percentage changes in the EMG burst parameters and in the peak amplitude of the EMG profiles (peak EMG) were calculated. EMGs were recorded from vastus lateralis (Vastus), biceps femoris (Biceps), semitendinosus (Semiten), gastrocnemius medialis (Gastroc), and soleus. '+' indicates increase, and '-' indicates reduction. In all cases the onset time was shifted towards the heel contact. * indicate that the changes were non-significant (P>0.05).

	Vastus	Biceps	Semiten	Gastroc	Soleus
Onset (%)	*	6.5	13.4	25.5	*
Duration (%)	-9.8	*	-5.9	*	*
RMS (%)	+18	+38	+21	*	*
MPF (%)	*	-7.8	-13	*	*
Peak EMG (%)	+28	+70	+38	+14	+10

The most pronounced changes in the muscle activity (RMS, MPF, peak EMG) and the coordination (onset, duration) occurred for the hamstring muscles (table 1). Gastrocnemius was recruited substantially closer (25%) to heel contact during fatigue, although no significant EMG amplitude changes were found. The burst RMS values were significantly increased for vastus lateralis, biceps femoris and semitendinosus; the largest increase (39 %) for biceps femoris. No burst onset and offset could be detected for tibialis anterior as it was active during the entire stride. The MPF decreased for the hamstrings muscles with a maximum of 12 Hz for semitendinosus (fig. 1). No changes in the pattern of the EMG profiles occurred during fatigue, but the amplitude modulation of the EMG peak values were found (table 1, fig. 2) for vastus lateralis (28%), biceps femoris (70%, fig. 2) and semitendinosus (38%).

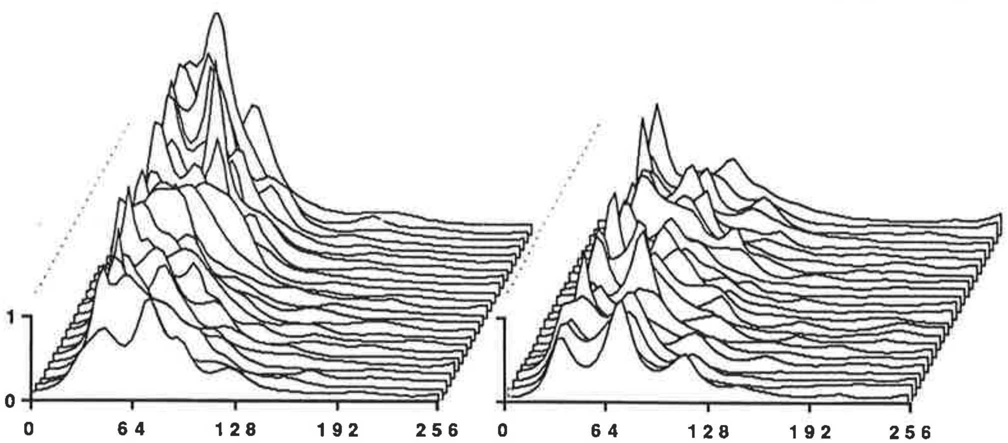

Fig. 1. Power spectra calculated for EMG burst activity in individual stride cycles for the biceps femoris (left) and the soleus (right) muscles. Each power spectrum represents the average of 6 steps. Horizontal axis in Hz.

DISCUSSION

The present study indicates that muscle coordination and muscle performance are affected during high intensity cyclic muscle load, but the pattern of the leg movements remained constant. If human locomotion is controlled by pattern generators, this system seems to have input which enables it to maintain the kinematic of e.g. the knee joint although the muscles are fatigued. Both the peak amplitude of the EMG profile and the onset time of the bursts changed more than the MPF, which is traditionally used to assess muscle fatigue. The MPF value is sensitive to most of the parameters that vary during dynamic contractions (e.g. firing rate, recruitment, muscle fiber conduction velocity, temperature, muscle length, contraction velocity) indicating that the MPF might not be adequate for the quantification of dynamic muscle fatigue. It has been shown that the MPF and the muscle fiber conduction velocity are also dependent on muscle tension. The MPFs for the individual EMG bursts represent a range of contraction levels, but may be dominated by the large amplitude, and the fast conducting motor units recruited at the highest contraction forces. During walking, the muscles are continuously shortened and extended. Shortened muscles are more easily to fatigued (2), which could be a reflection of an inability to activate the motor neuron pool at the higher rates required to achieve tetanic fusion for each motor unit when the muscle is shortened (3). The MPF increases as the muscle length decreases (4), but this does not seem to influence our results. The actual length of the hamstring muscles during walking is difficult to evaluate as they are biarticulate muscles. For constant tension and contraction velocity concentric contractions produce greater EMG than eccentric contractions, and furthermore the EMG is dependent on the speed of contraction (5). As the EMG profile amplitude is modulated by many parameters, the amplitude can therefore not simply be related to the actual tension as for static contractions.

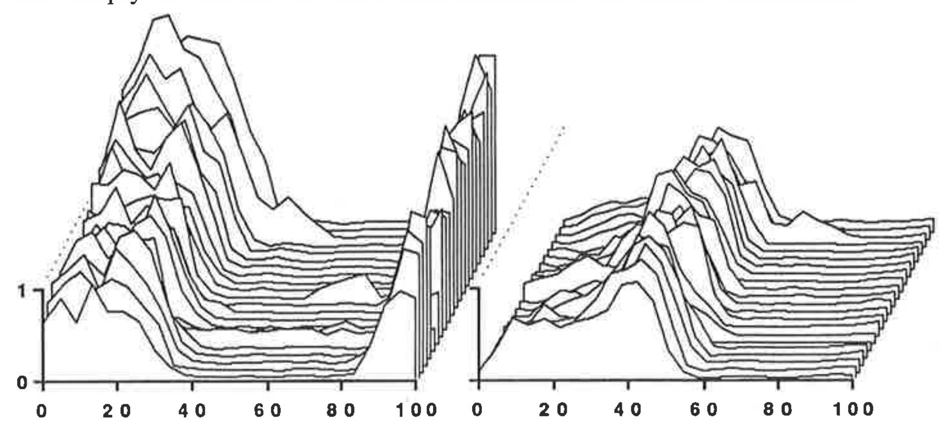

Fig. 2. Pseudo three-dimensional plots of single EMG profiles from the biceps femoris (left) and the soleus (right) muscles, in which the heel strike is 100%. Each profile represents the average of 6 steps.

The intramuscular temperature increases during work and the MPF increases with increasing muscle temperature (6). The minor decreases in the MPF found during uphill-walking, might therefore be larger than what was actually measured. It is known that the muscle blood perfusion is important for the development of the EMG characteristics during fatigue. During walking, the intramuscular pressure is occasionally sufficient to cause intramuscular ischemia (7), but normally these periods are followed by periods with restored circulation.

Our findings that no EMG changes occurred in the soleus and that dramatic changes occurred in the hamstring muscles may be explained by different fiber compositions, of which the soleus is known to be a slowly contracting, fatigue-resistant muscle. The changes in integrated EMG and MPF during maximum fatiguing knee extensions are previously found to be correlated with the fiber composition (8). Whenever dynamic muscle fatigue is studied, it may be a problem to obtain a sufficient degree of muscle fatigue and at the same time not to exhaust the cardiovascular system. In work physiology, it is essential to find adequate techniques to quantify dynamic muscle fatigue as it will provide better possibilities to plan appropriate postures for e.g assembly workers who perform repetitive muscle work of low intensity.

ACKNOWLEDGEMENTS

Det Obelske Familiefond, Aalborg Stifts Julelotteri and Assurandør Societetet were kindly acknowledged for their financial support.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

QUANTITATIVE ANALYSIS OF MUSCLE FATIGUE BY PARAMETRIC POWER SPECTRUM OF NONSTATIONARY ELECTROMYOGRAPHY DURING A QUICK MOVEMENT

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INTRODUCTION

Since surface electromyography (EMG) signal represents electrophysiological activity in human intact muscles, it has been used as means of evaluation of the muscle activity for various contractions. Many studies have reported an amplitude increase and a frequency lowering of the EMG signal in fatigue state during isometric contraction(1)(2). However, investigation for muscle fatigue during a quick movement of upper limb has been rarely shown. Dynamic EMG signal of muscle during a quick movement shows nonstationarity,and so it is not easy to analyze muscle fatigue state by using the EMG signals. Power spectrum analysis has been widely used as a means of data reduction and characterization of EMG signal. In particular,parametric power spectrum method by using the autoregressive model has been proposed as a method which can eliminate the drawbacks of nonparametric method(FFT) such as spectral line splitting, frequency bias, and also can process signals of very short data block(3). In this study,we propose the use of parametric power spectrum method used optimum weighting function for processing the nonstationary EMG signals in fatigue state. By using %median frequency and %power below the initial median frequency in power spectrum of the dynamic EMG signals, the muscle fatigue state for various muscles during fatiguing contractions is analyzed quantitatively. Also, we evaluate the change of blood lactate concentration as an important tool which explain phyiologically the cause of spectrum shift due to fatigue state.

MATERIAL AND METHODS

Parametric power spectrum method

The computational techiques to analyse muscle fagiue state is expanded as shown in *Fig.1*. Once the linear prediction error filter is implemented by using the lattice filter structure, then the relationship between forward [$e_f(P,t)$] and backward [$e_b(P,t)$] autoregressive models(F-B AR model) for dynamic EMG signals X (t) are established as eq.(1),(2);

$$e_f(P,t)= e_f(P-1,t)+ a(P,P) \cdot e_b(P-1,t-1) \qquad (1)$$

$$e_b(P,t)= e_b(P-1,t)+ a(P,P) \cdot e_f(P-1,t-1) \qquad (2)$$

where P indicates model order(=6).

The initial condition of model is defined as $e_f(0,t)= e_b(0,t)= X(t)$.

And,AR model parameter a (P,i) is obtained by eq.(3)

$$a(P,i)= a(P-1,i)+ a(P,P)\, a(P-1,P-i), \qquad a(P,P)=N(t)/\sqrt{\ } D(t) \qquad (3)$$

$$N(t)=-\sum_{t=P}^{N-1} W(P-1,t)\, e_f(P-1,t)\, e_b(P-1,t-1)$$

$$D(t)=[\sum_{t=P}^{N-1}(e_f(P-1,t))^2\sum_{t=P}^{N-1}(e_b(P-1,t-1))^2]$$

where a (P,P) is reflection coefficient which is called as correlation coefficient between forward and backward predictor, and W(P-1,t) indicates a nonnegative optimum weighting function, N indicates data length. The optimum weighting function W(P,t) is given as eq.(4);

$$W(P,t)=2W(P,t-1)-W(P,t-2)-F(P),\ F(P)=12/(N-P+1)(N-P+2)(N-P+3) \qquad (4)$$

where F(P) is Lagrange's multiplier, and F(0)=0 and F(1)=NF(1)/2.
By using the optimized value of F-B prediction error variance V_p and the identified AR parameter, parametric power spectrum is represented by the form of maximum entropy method as eq.(5);

$$P(f)=\frac{V_p\cdot D}{|1+Ex(f)|^2},\quad Ex(f)=\sum_{i=0}^{P}a(P,i)\cdot exp(-j2\pi fiD) \qquad (5)$$

where D indicates the sampling rate(=0.4ms).

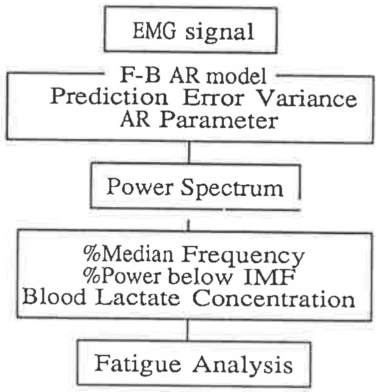

Fig.1 Computational techniques

Fig.2 Experimental system

Experimental system and method

Fig.2 shows the experimental system composed of mechanical, EMG and metabolic system. Six normal subjects were studied. Surface electrodes of bipolar type with 8mm in diameter were placed on the surface of primary muscle (brachioradialis,biceps,tricpes). We concentrated on repeated quick movement of the upper limb for maximum contraction level(more than 50%MVC), angular velocity(100deg/sec) and angle displacement(30deg). The subjects were sat in a chair of dynamometer(Chattex Co.,KIN-COM Ⅲ) which measures the mechanical parameters such as the angular velocity , the force and the angle change simultaneously. The dynamic surface EMG signals were differentially amplified, lowpass filtered, and were transmitted to computer through A/D converter with 2.5KHz of sampling frequency. To consider the changes of intramuscular related to EMG spectrum shift,the lactate concentration in blood was measured. Blood samples were drawn from the forearm vein over before exercise and among quick contractions of 3minutes and after exercise. Lactate component in blood was analyzed by the lactate analyzer(YSI Co.Model 231L).

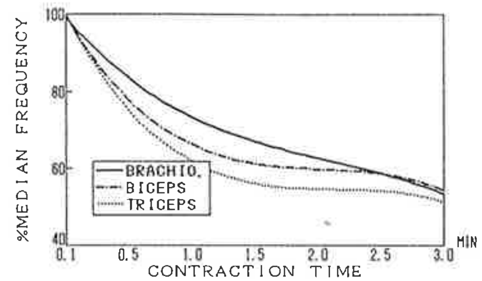

Fig.3 Time trends of %median frequency of EMG signals during a quick movement for maximum contraction level

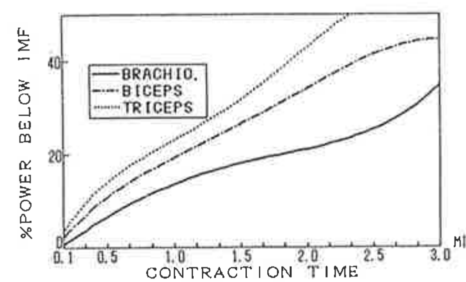

Fig.4 Time trends of %power below IMF in EMG power spectrum

RESULTS AND DISCUSSION

The data presented below were obtained exclusively from 3 minutes exercise. *Fig.3* depicts time trends of %median frequency of the dynamic EMG signals for various muscles during repeated quick movement of upper limb under maximum contraction levels. In this figure,it could be shown that %median frequency as compared to the initial value before muscle fatigue was linearly decreased until 58%. *Fig.4* shows changes of %power below initial median frequency(IMF) in EMG spectrum of the brachioradialis, the biceps and the triceps muscle for repeated quick movement. This result suggested how the cumulative power below initial median frequency for various muscles during fatigue state was changed. That is, time trends of %power below initial median frequency represent the changes of EMG amplitude as well as a frequency lowering caused by muscle fatigue.In particular, during contraction time point of 2 min ,it was observed that %power below IMF in the brachioradialis,the biceps and the triceps muscle increased until 21%, 35%, and 44%, respectively. It might be explained that each muscles has different fatiguability and this difference is influenced by structure and kinds of muscle fibers in primary muscle relating to repeated quick movement. *Fig.5* shows changes in blood lactate concentration drawn from the forearm vein before exercise(rest state), during repeated quick movement (contraction) of 3minutes and after exercise(recovery state). In this figure, we can find that changes of lactate concentration were influenced on degree of a contraction level. Lactate concentration for 25%MVC and 50%MVC were elevated to 3.6mmol/l and 4.5mmol/l with the lapse of 2 minute after exercise, respectively, while peak concentration for 10%MVC was increased no more than 1.5mmol/l in just after exercise of 3 minute. Finally, in order to investigate interrelation between %median frequency and %power below IMF and blood lactate concentration,regression analysis was applied for maximal contraction level as shown in *Fig.6(a)(b)*.It could be confirmed that regression analysis shows a significant correlation that a coefficient of correlation between %median frequency and blood lactate concentration for repeated quick movement of 3 minitues is -0.9530($p<0.01$,T-test) and between %power and blood lactate is 0.9739($p<0.01$).

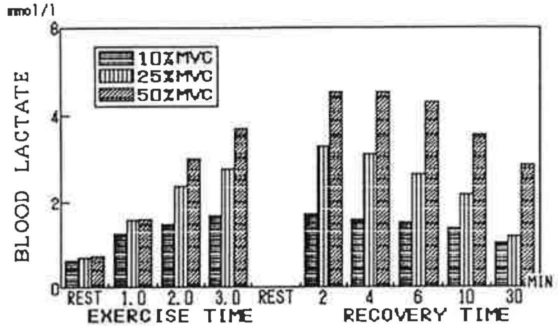

Fig.5 Time trends of blood lactate concentration before exercise and during repeated quick movement(3 minutes) and after exercise(recovery interval)

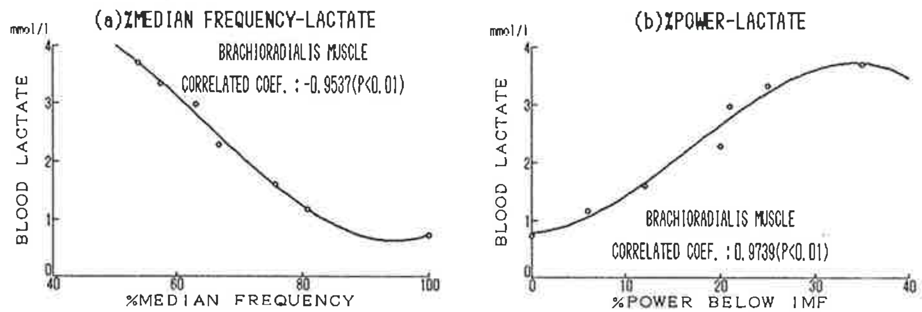

Fig.6 The relationships(a)between %median frequency and lactate concentration (b)%power below initial median frequency and lactate concentration

In summary,by using the proposed method and parameters, we have been able to bring into evidence that the phenomena of spectrum change of nonstationary EMG signals during repeated quick movement of upper limb have a physiological significance. Also it would be possible to estimate that spectrum changes during muscle fatigue correlate with blood lactate concentration.

In conclusions, changes of EMG power spectrum due to muscle fatigue during repeated quick movements of upper limb function were analyzed as (a)shift of median frequency to low region (b)increase in %power below initial median frequency(c)different fatiguability of each muscles relating to contractions (d)a siginficant correlation between changes in the lactate concentration in blood and the spectrum changes due to fatigue.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

CROSS-SPECTRUM OF MUSCULAR SOUND AND SURFACE EMG. A TOOL TO MONITOR FATIGUE AT MAXIMAL VOLUNTARY CONTRACTION

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INTRODUCTION

During one minute of sustained maximal voluntary contraction (MVC) a well described reduction of about 50% of the mean motor units (MUs) firing rate takes place. The above reported phenomenon was described in different human muscles such as the abductor pollicis (2) and the biceps brachii (4). The same MUs firing pattern information is contained, even if masked by the interferential nature of the two signals, in the surface EMG (SEMG) and in muscular sound or sound myogram (SMG). This last possibility was suggested by Gordon and Holbourn (3) which in their work detected, by means of a piezoelectric crystal, one sound spike for each motor unit action potential recorded.

On these basis the aim of this work is to verify if it is possible to retrieve, by the use of the cross-spectrum analysis technique applied to the SMG and SEMG, the changes in the mean MUs firing rate, that take place during fatiguing MVC.

MATERIALS AND METHODS

Ten healthy sedentary males (age: 22-24) volunteered for the study. For SMG detection a contact sensor transducer (H.P. 21050A, bandwidth: 0.02-2000 Hz) was placed over the belly of the biceps brachii and firmly strapped, by an elastic band, to the muscle. For EMG recordings, two surface electrodes, center to center distance 1 cm, were placed close to the piezoelectric sensor. The force generated by the flexors of the elbow was measured by a load cell strapped perpendicularly to the subject's wrist. A visual feedback of the developed force was provided by means of an oscilloscope. During the effort the subjects' arm and forearm were kept in an anatomically shaped stirrup allowing a 115^{0} between the two elbow joint segments. Throughout the exercise the subject was asked and encouraged to develop the maximum output force by his elbow flexors muscles until exhaustion was reached. Before A/D conversion the SEMG and SMG were low pass filtered at 500 and 120 Hz, respectively. Force, SEMG and SMG were sampled at 1024 p.p.s. and stored on the hard disk of an IBM-AT computer. The final format in which the data were stored was a sequence of 21 epochs of 2 s each, from 0% to 100% of the total contraction time (TCT) in step of 5% TCT. On the 2 s epochs after the EMG and SMG auto spectra calculation, by means of the FFT with a Popoulis window, the cross-spectra between the two signals were estimated. For both spectra and cross-spectra their mean frequency (MF) was calculated.

RESULTS

The average duration of the exercise was 83±8s (mean ±SE). During this time the force reduced exponentially from 311±10 N to 88±8 N with a halftime of about 20 s.

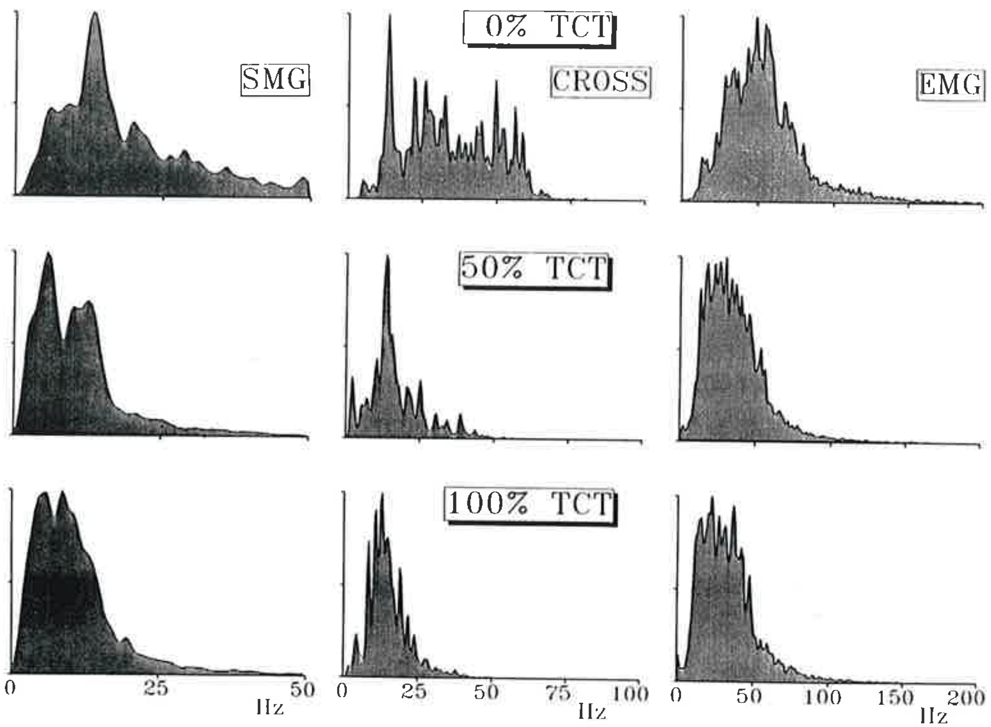

Fig. 1 - Average of the normalized SMG and SEMG auto and cross-spectra at 0%, 50% and 100% of the total contraction time (TCT).

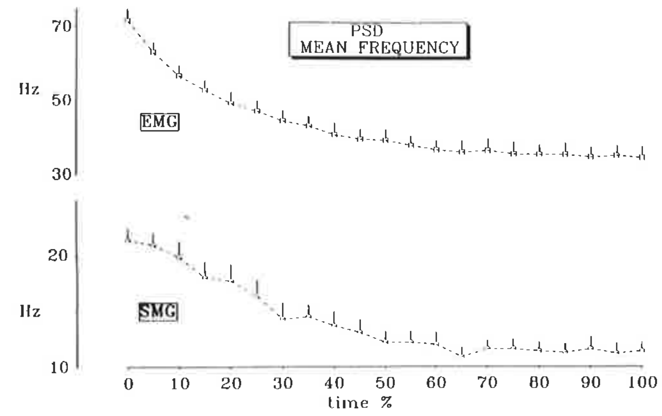

Fig. 2 - Average values (+SE) of the SEMG and SMG power spectrum density distribution (PSD) mean frequency as a function of time.

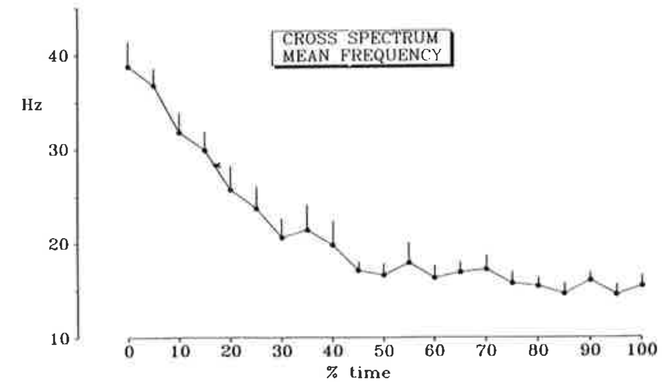

Fig. 3 - Average values (+SE) of the SMG and SEMG cross-spectrum MF as a function of time.

As reported in Fig. 1 it can be noted that throughout contraction the auto and the cross-spectra of SMG and SEMG became narrower, mainly because of a loss of power in the high frequency bandwidth, and a shift toward the low frequencies. The described changes in the morphology and in the position of the spectra with time are clearly reflected in the MFs trend. In fact, from the onset to the end of the effort the MF of the SMG and SEMG reduces from 21±1 Hz and 71±4 Hz to 11±1 Hz and 35±3 Hz, respectively (Fig. 2). The changes in the average value of the cross-spectrum MF is reported in Fig. 3. It decreases from 37±2 Hz to 15±2 Hz in a near exponential fashion, with a halftime of about 20 s.

DISCUSSION

During sustained MVC of about 60 s the mean motor units firing rate (FR) could be reduced of about 50% and the histogram of the relative number of MUs vs FR shifts to the left and becomes narrow (2). This reduction in the mean MUs frequency rate was related to the increase of the muscle fibres relaxation time during fatiguing MVC, which could influence the activity of the motor neurons via a "reflex mechanism" (2). The cross-spectrum (CS) between SMG and EMG seems to reflect the above described phenomena. In fact its bandwidth (about 10 ÷ 60 Hz) at the onset of the MVC is close to the MUs vs FR histogram reported for the biceps brachii and the abductor pollicis in the same experimental conditions (1). The changes in CS shape and position with time during sustained MVC fit well with the MUs-FR histogram characteristics of the fatigued muscle (bandwidth: 5÷25 Hz) (2). At the same time the exponential reduction of about 50% of the MF of the CS is comparable with the reduction of mean MUs' FR from the onset to the end of the 90 s sustained MVC (2).

In conclusion, the cross spectrum of SMG and SEMG can be considered a useful and non-invasive tool to monitor the influence of the localized muscle fatigue on the motor units firing pattern during sustained MVC.

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ADAPTATION TO FATIGUE OF LONG DURATION IN HUMAN WRIST MOVEMENTS

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INTRODUCTION

Muscle fatigue can result from a deficit in one of three mechanisms: excitation, contraction, or the coupling between the two. Fatigue of the coupling process between electrical excitation and mechanical contraction occurs even during moderate exercise and recovers only very slowly, persisting for up to twenty-four hours (1). Underlying the decrease in muscle force production is a depressed muscle twitch (2). A decrease in the amount of calcium released from the sarcoplasmic reticulum is likely to be responsible for decreased force production in long term fatigue (3). This is thought to be due to a failing link in signal transduction from the transverse tubules to the sarcoplasmic reticulum (4), perhaps from a derangement of the transverse tubules, themselves (5). In humans, the clinical and practical consequences of fatigue of long duration may be significant. Although the cause(s) of long term fatigue are fairly well understood, this is the first report concerning adaptation to this condition.

MATERIALS AND METHODS

Movement Task Description

Subjects were seated at a table with their right forearm secured between wooden blocks. An oscilloscope placed in front of the subject displayed a vertical line representing hand position. The "target" for the task consisted of two lines separated by a narrow margin (corresponding to 0.2 degree). Starting from close to anatomical position, the subject was instructed to "move as quickly as possible to the target". After much practice, fifty 30-degree movements were recorded.

Fatiguing Protocol and Twitch Recording

Subjects then fatigued wrist flexors and extensors by lifting weights. Exercise continued until muscle twitches in response to stimulation of the ulnar nerve declined to approximately 30%

of their original force. Subjects then rested one hour in order to allow restoration of maximum voluntary contraction, but not twitches. Following rest, twitches were again recorded to ensure depression from fresh values. The subject then performed the movement task: fifty trials made "as fast as possible" under fatigued conditions.

Data Recording and Analysis

Position, acceleration, torque, and flexor and extensor EMG's (flexor carpi radialis and extensor carpi ulnaris) were simultaneously sampled 500 times per second and recorded for the course of the movement. Movements were later synchronized at the time of peak velocity and by overlaying synchronized acceleration traces on a graphical display, we were able to select ten similar trials for both normal and fatigued conditions. The twenty trials with nearly identical dynamics were used for comparison of EMG traces.

RESULTS

Fatigue of long duration is characterized by depressed muscle twitches. Twitch tensions recorded before (T1), immediately after (T2), and one hour following fatiguing exercise (T3) are listed for four subjects in Table 1. Twitches are also given as a percentage of fresh value (%T1).

TABLE 1

Twitch Tension (N)

Subjects	S1	S2	S3	S4
T1	9.16	12.31	18.49	8.70
T2	4.08	3.29	5.52	3.54
%T1	44.54	26.75	29.86	40.70
T3	4.55	4.47	10.53	5.27
%T1	49.68	36.31	56.92	60.62

Even though muscle twitches had recovered to only approximately 50% of fresh values, subjects performed the same movement consistently under normal and fatigued conditions. The average position, acceleration and torque for ten movements under normal and fatigued conditions are shown for one subject (Figure 1). The trajectories are nearly indistinguishable,

although muscle twitches have only recovered to approximately 60% of fresh value. There must then be a difference in the control pattern sent to the muscle in order to compensate for the fatigued condition.

EMG's recorded during the movement differ under fatigued conditions in a few main respects: 1) the onset of the first agonist burst is earlier (with respect to peak velocity) 2) the width of the first agonist burst is increased and 3) the area of the second agonist burst is increased with fatigue.

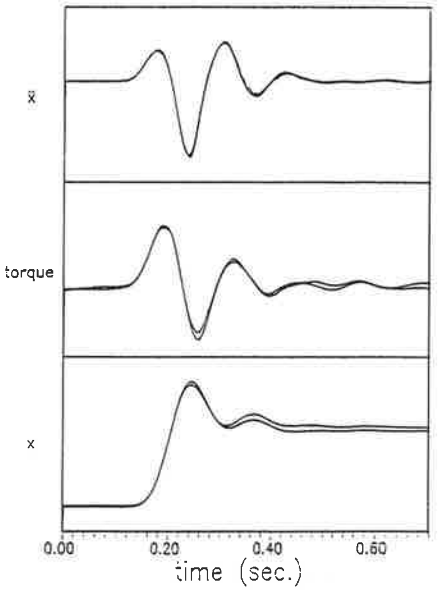


Fig. 1. Average acceleration, torque, and position traces performed under normal and fatigued conditions are nearly indistinguishable.

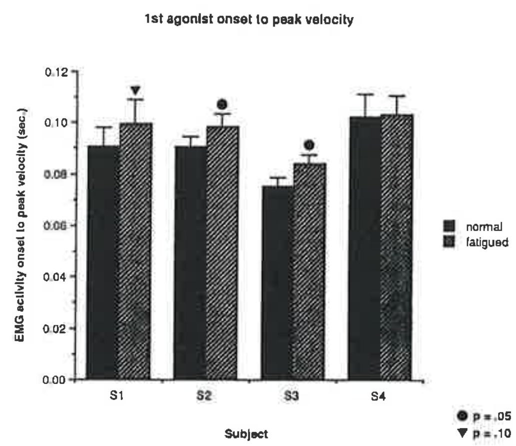


Fig. 2. The time from the onset of agonist EMG activity to the point of synchronization (peak velocity) is longer in fatigue.

Figure 2 depicts the earlier onset of the first agonist with reference to the point of synchronization (peak velocity). In all subjects, the first agonist is activated earlier in fatigue. In Figure 3, agonist pulse widths are plotted for four subjects during both conditions. "Pulse width" is defined from the initial onset of the burst to the time at which the smoothed, rectified burst has decreased to half its maximum value. Lastly, subjects appear to compensate for the fatigued condition by increasing the area of the second agonist burst (Figure 4).

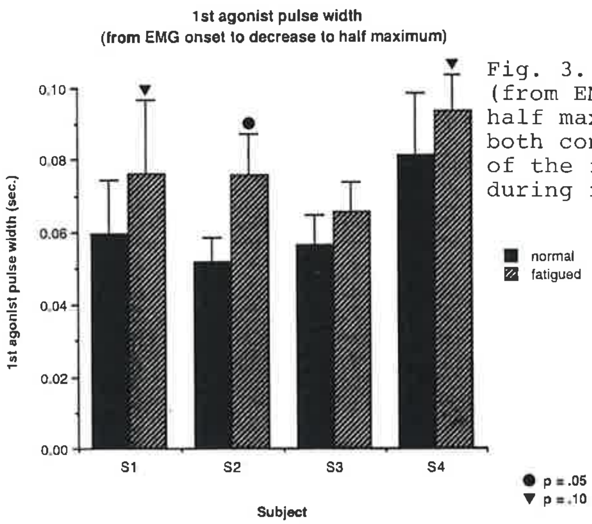
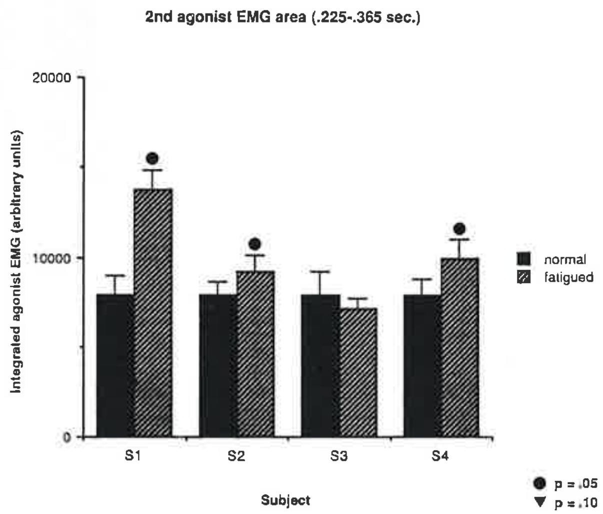


Fig. 3. Agonist "pulse widths" (from EMG onset to decrease to half maximum) are plotted for both conditions. The duration of the first agonist is longer during fatigue.

Fig. 4. Agonist EMG activity,integrated over a time interval covering the second burst, is plotted for all subjects under both conditions. The area significantly increases in fatigue.



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Electromyographical Kinesiology
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TEMPERATURE EFFECTS ON EMG AND MUSCLE FUNCTION

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INTRODUCTION

The temperature of limb muscles can vary widely as a result of the environmental temperature and the level of activity. It is reported that changing the temperature of the muscle results in changes in muscle function (2, 5), and in the electromyographic activity of the muscle (13). However, most of these studies only investigated static muscle contraction at force levels above 20-30% of the maximal force of the muscle. These force levels are expected to be above the force levels used in most daily live activities.

The aim of this study was to investigate the relationship of EMG and muscle function at relatively low dynamic and static contractions of the forearm muscles at different temperatures.

METHODS AND MATERIALS

Subjects

Nine young male volunteers (age 18-24 years, weight 70-82 kg, height 170-186 cm) agreed to participate, after being fully informed about the aim and the procedures of the experiment.

Muscle function

The isometric strength was measured on a custom built handgrip dynamometer. The maximum isometric strength was assessed as the largest of three brief, maximal voluntary contractions (MVC). Maximal rhythmic grip frequency was used as dynamic exercise. The subjects were instructed to perform rhythmic contractions between 0 and 10% MVC (dependent of the temperature condition) for a 30 s period. During the experiment the subjects were seated with their right arm in a water bath, with an elbow angle of 90°. Also, the maximal endurance time was determined at 15% MVC.

EMG

The surface EMG was recorded with two electrodes (Ag-AgCL, Ø 9 mm), placed 2 cm apart over the belly of the m. flexor digitorum superficialis. A guard electrode was placed on the bony surface of the acromion. The EMG was amplified and A/D converted with a sample frequency of 2,048 Hz. The RMS amplitude of the EMG was measured with a custom built RMS detector. The power spectrum was calculated, with a time interval of 1 min, for a 1 s period by means of fast Fourier transform. As a parameter for description of the power spectrum the median power frequency (MPF) was taken, the frequency above and below which the integrated power is equal.

Temperature control

The temperature of the forearm was varied by immersion in a water bath for a period of 30 min. The water temperature was set at either 15 or 40 °C, resulting in corresponding skin temperatures. A reference condition of a skin temperature of 32 °C was attained in air (24 °C).

Experimental procedures

In each temperature condition the MVC was determined. The subjects were then asked to perform brief contractions at 5, 10, 15, 20, 50, 90% MVC with an interval of 1 min. After 5 min of rest the maximal rhythmic grip frequency was determined for a period of 30 s. Finally, after a rest period of 15 min, the subjects performed a sustained contraction at 15% MVC.

RESULTS

Static exercise

The average maximal grip strength was with 326 N significantly reduced in the coldest situation (15 °C) compared to the 32 °C (417 N) and 40 °C (382 N) condition.
The relation between RMS and the exerted force remained linear and was not significantly influenced by the temperature.

Dynamic contraction

Cooling the arm to 15 °C significantly reduces the maximal grip frequency with 50% (Fig. 1). An arm temperature of 40 °C did not significantly influence the maximal grip frequency compared to the 32 °C temperature condition.

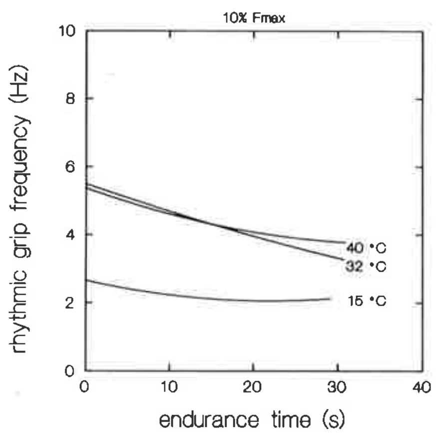

Fig. 1. The effect of temperature on the maximal rhythmic grip frequency (Hz) for a 30 s exercise period.

Sustained contraction

The endurance time for a contraction at 15% MVC (dependent on the temperature) was with 60% significantly reduced in the 40 °C condition, compared to the 15 and 32 °C skin temperature condition. Cooling the muscle did not influenced the endurance time compared to the 32 °C condition. The increase in the RMS with contraction time was also dependent on

the temperature condition (Fig. 2). In the 40 °C condition the average RMS value at the end of the contraction was 200% of the initial value. However, in the 32 °C condition the final RMS was 300% and in the 15 °C condition the final RMS was only 144% of the initial value.
The MPF did not decrease significantly with time in the 40 and 32 °C temperature condition (Fig. 3). In the 15 °C temperature condition the MPF started at a significantly lower value, but increased with contraction time to the value of the 32 °C temperature condition.

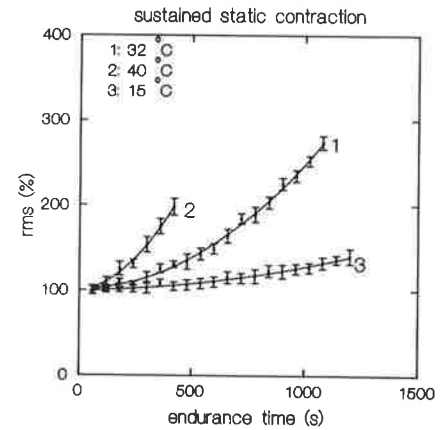

Fig. 2. The increase in the RMS (% mean ± SE) of the EMG with contraction time (s), dependent on the temperature condition.

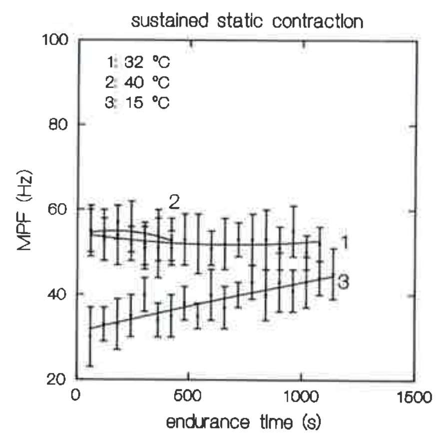

Fig. 3. The MPF (mean ± SE) during a sustained contraction at 15% MVC, dependent on the temperature condition.

DISCUSSION

The results of this study, indicating a significant reduction of the maximal static strength due to cooling of a muscle, are in agreement with the effects reported in literature (1,6). It is hypothesized in literature that cooling a muscle below a temperature of about 27 °C interfere with neuro- and neuromuscular transmission of the superficial muscle fibres, resulting in a reduction of the force output (5, 7). Most of studies also report a reduction in endurance time of sustained contractions with high muscle temperatures (5, 8, 13). It is thought that this effect is the result of an enhanced glycolysis in the heat (3, 9, 10) leading to an accumulating of metabolic endproducts, which partially inhibit the phosphorylative efficiency of the muscle mitochondria (4, 9). The effects of temperature found in this study on the parameters measured during the sustained contraction indicate that the MPF and RMS values do not unambiguously reflect the decrease in endurance time during the 40 °C temperature condition. The amount of increase in the RMS amplitude found during a sustained contraction is reported to be increasing with an increasing level of fatigue (14, 12). However, in this study the amount of RMS increase in the 40 °C temperature condition is less than the amount of increase in the 32 °C temperature condition, despite the shorter endurance time in the former condition. Also, the normally reported decrease of the MPF with fatigue (11, 13) was not found in the hot condition.

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EFFECTS OF BATH pH ON EMG MEDIAN FREQUENCY AND CONDUCTION VELOCITY IN THE HAMSTER DIAPHRAGM

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Introduction

The physiological determinants of EMG fatigue parameters have not been clearly established. H^+ ion accumulation at the sarcolemma is believed to play a key role in determining the electrophysiological correlates of fatigue. To establish a causal relationship between muscle pH and EMG, studies must go beyond merely recording concurrent changes in these measures. Concurrent recordings of metabolic and electrophysiologic changes only describe associative relationships because this approach is limited by the inability to distinguish between biochemical changes that cause the observed EMG changes, and those that do not effect the EMG signal. An in-vitro model provides a direct method of establishing causality because an isolated change can be induced and the resultant EMG changes recorded.

The purpose of this study was to selectively alter the pH of a nerve-muscle preparation and record the resultant alterations in the EMG signal. Specifically, we determined the effects of bath pH on the Median Frequency (MDF) and Conduction Velocity (CV) recorded from an isolated diaphragm. We also addressed the question of whether the relationship between MDF and CV is different during induced changes in bath pH compared to changes that occur during stimulated contractions which fatigue the muscle preparation. This analysis was conducted to determine if the changes in MDF that occur during fatigue are the result of more than just changes in muscle pH.

Materials and Methods

Animal Preparation: Eight adult LVG Syrian Golden hamsters were anesthetized via pentobarbital sodium (6 mg/100 gm bw). A tracheostomy was performed and the left hemidiaphragm with intact phrenic nerve was removed. A 5 mm wide costal diaphragm strip was isolated using a template and its tendons placed between two plexiglass clamps. The neuromuscular preparation was then placed in a jacketed bath containing Krebs solution maintained at 26 °C by a circulating pump. The solution was continuously aerated with a mixture of 95% O_2 and 5% CO_2 which maintained the bath pH at 7.4. The unattached end of the nerve was hooked to a bipolar

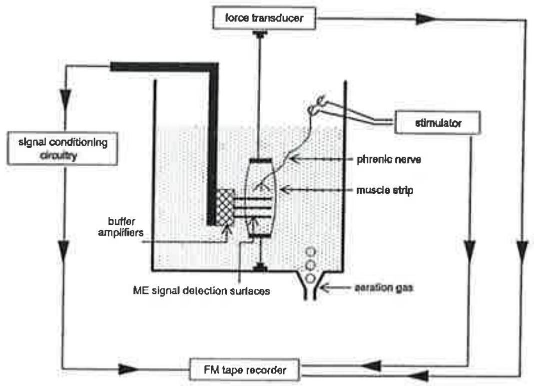

Figure 1: The experimental set-up

stimulating electrode, and a supermaximal voltage was determined; the stimulating voltage was set at 1.5 times this value. The muscle length was then adjusted to its optimal length, defined as the length at which peak twitch amplitude could be generated. The supermaximal electrical impulses were delivered at 40 Hz (0.2 ms pulse duration) for 3 seconds through a stimulation isolation unit. The complete set up of the experiment may be seen in Figure 1.

Experimental Procedures: Experiments were designed to evaluate the effects of bath pH changes on the EMG signal MDF and CV. Each strip was exposed to normal bath conditions (bath pH 7.4) and two levels of respiratory acidosis (bath pH 7.0 and 6.6). Decreases in bath pH were achieved by increasing the CO_2 concentration of the aerating gas mixture. Solutions at these pH levels were previously equilibrated, and pH changes were achieved by a drain and replace technique. The sequence of pH exposure was randomized to either 7.4,7.0,6.6 or 6.6,7.0,7.4. Three repeated contractions were performed at each bath pH level, allowing a 5 minute rest period between successive contractions. A 15 minute equilibration period between bath pH changes was allowed to ensure a static relationship between bath pH and resting intracellular pH. These equilibrations period were shown to be sufficient by us in preliminary experiments by noting stable values of tension, MDF and CV.

ME Signal Detection Technique: A special electrode probe with three detection bars was developed for differential detection of EMG activity. The detection bars were made of compliant gold-plated springs, and the inter-electrode spacing was 2.38 mm. The detection array was placed so that all three bars were located on the distal side of the motor end-plate region. Signal conditioning circuitry was located both inside and outside of the bath. High impedance buffer amplifiers (gain =1) were connected directly to each detection surface in the bath. Insulated cable connected the buffers to a circuitry box located near the bath. Functionally, the circuitry box consists of an instrumentation amplifier (differential gain =10), a band pass filter (20 - 2000 Hz), and an adjustable gain (range 10-500). The conditioning circuitry provided three differential EMG signals. One of these signals was used to compute MDF by implementing an FFT algorithm over an averaged M-wave representing each 0.5 s of data. The other two signals were used to estimate average CV by comparing the phase shift using a cross-correlation technique (further details described in (5)).

Results and Discussion

Typical MDF vs time and CV vs time plots for one preparation are presented in Figure 2. In general, the initial values of both parameters were highly repeatable for the 3 consecutive contractions at each bath pH. There were no cases in which the three initial values of MDF or CV varied around their mean by more than 3%. Initial values of MDF and CV are defined as

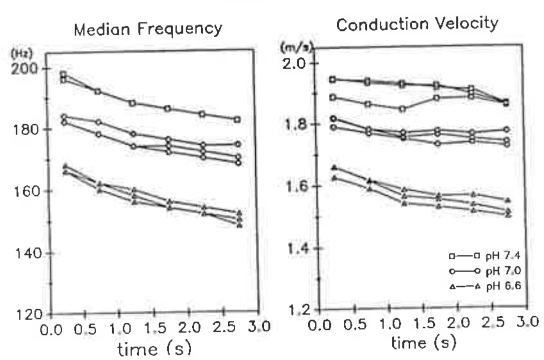

Figure 2: Typical time course of MDF and CV

values calculated at t= 0.25 s and final values of MDF and CV are defined at t= 2.75 s

Effects of Bath pH on the Initial Values of MDF and CV: The mean initial MDF (IMDF) vs bath pH and mean initial CV (ICV) vs bath pH are shown in Figure 3. The results were similar regardless of the ordering of pH changes. In each preparation, the initial value was normalized with respect to the average of the 3 initial values at pH 7.4. The mean IMDF at bath pH 7.4 was 1.00 ± .008 and the mean initial CV was 1.00 ± .012. Decreasing the bath pH resulted in decreases in both IMDF and CV in each of the 8 preparations. The mean normalized IMDF at bath pH 7.0 was .948 ± .027, and was .854 ± .029 at bath pH 6.6. The mean normalized ICV at bath pH 7.0 was .947 ± .033, and was .863 ± .035 at bath pH 6.6. The differences in the initial values (both IMDF and ICV) at different pHs were significant ($p < .01$) in all cases. In general, the bath pH change resulted in an equal percent change in IMDF and ICV. There was no statistically significant difference between normalized initial MDF and initial CV values at pH 7.0 or pH 6.6.

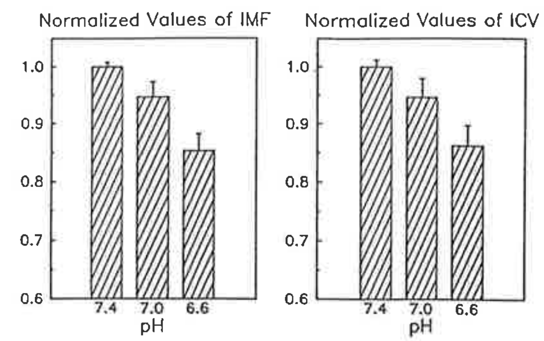

Figure 3: Normalized values of IMDF and ICV at different bath pH's

CV acts as a linear operator on the action potential (3). When the CV of an action potential is scaled, its spectrum is scaled by the same factor. This relationship holds true for the spectrum of the M-wave, although noise is effectively added to the relationship because the spectrum of the M-wave is produced by a non-linear summation of the action potential spectra. Therefore equal percent changes in MDF and CV implies that the spectral changes observed during bath pH changes were merely due to the induced CV changes, with no change in the fundamental shape of the M-wave.

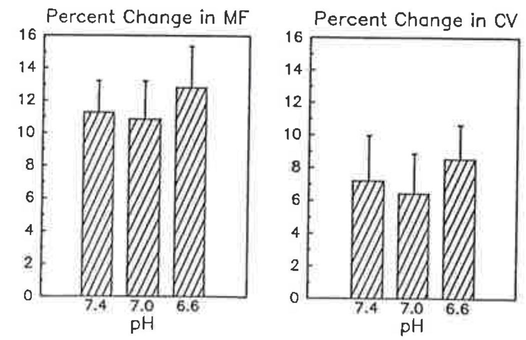

Figure 4: Percent change in MDF and CV during stimulated contractions

Changes in MDF and CV During Contractions: Mean percent changes of MDF and CV during contraction are presented versus bath pH in Figure 4. Percent change is defined as:

$$Percent\ Change = \frac{(Initial\ Value - Final\ Value)\text{x}100\%}{InitialValue}$$

At bath pH 7.4, MDF decreased by 11.25 ± 1.95%, and CV decreased by 7.20 ± 2.77%. At bath pH 7.0, MDF decreased by 10.86 ± 2.35%, and CV decreased by 6.43 ± 2.44%. At bath pH 6.6, MDF decreased by 12.80 ± 2.56%, and CV decreased by 8.54 ± 2.09%. The mean percent decrease in MDF was significantly greater than the mean percent decrease in CV at all three bath pH's ($p < .01$). On the average, the rate of decay of CV was only 65% that of MDF. This implies that the spectral changes observed during sustained stimulated contractions were due to **both** changes in CV and changes in the fundamental shape of the M-wave.

Conclusions

1. Decreases in bath pH cause equal percentage decreases in Median Frequency and Conduction Velocity. The spectral changes are due to changes in Conduction Velocity.
2. During three second stimulated contractions Median Frequency decreases by a greater percentage than Conduction Velocity. The spectral changes are due to **both** changes in Conduction Velocity and to changes in the fundamental shape of the M-wave.

Acknowledgements
Supported by the Rehabilitation Research and Development Service of the Veterans Affairs and Liberty Mutual Insurance Company.

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MUSCLE REFLEX ACTIVITIES DURING FATIGUE

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INTRODUCTION

Among the different components of an electromyographic (EMG) response to a muscle stretch, it is agreed that the short latency component (SL) results from spindle monosynaptic Ia afferents on the motoneurons (MNs) (1). Our understanding of the longer latency EMG responses remains an open question and it has been recently suggested that they originate from other receptors such as cutaneous (2, 3) or cutaneous and joint receptors (4).

The study of muscle fatigue was found to be an interesting approach for the discussion of muscle contraction (5, 6) and SL reflex EMG component (7), but little is known about the effects of fatigue on the longer latency reflex EMG activity. This activity is considered as a whole (2, 8) or as being composed of at least by a medial latency component (ML) and a long latency component (LL) (9).

The present contribution compares the effects of fatigue during voluntary and electrically evoked contractions on SL, ML and LL reflex EMG responses to a quick stretch of the human first dorsal interosseous (FDI).

METHODS

The results were recorded from 27 healthy subjects of both sexes who gave informed consent to participate in this experimentation.

The mechanical stretches were ramp-and-hold movements. During the recording, the subject was required to exert an abduction torque of 10 % of his maximal voluntary isometric contraction against the stretcher (2). In general 16 to 32 EMG responses were averaged after recording by 2 surface electrodes fixed over the muscle motor point and the metacarpophalangeal joint of the index.

Muscle fatigue was induced either by a sustained maximal voluntary contraction or by an electrically evoked one at 30 Hz and the contractions were interrupted when the force fall to 50 % of its maximum value. The reflex EMG activities were recorded before and after the fatigue tests and every 5 min during the recovery for at least 15 min.

Muscle surface action potential (SAP) was obtained by delivering supramaximal pulses of 0.2 ms duration to the ulnar nerve at the wrist via two steel needle electrodes (6).

RESULTS

In response to a quick stretch, the FDI showed a first EMG activity of short latency (30.9 ± 3.0 ms; mean ± SD; n = 22) and a second EMG complex of longer duration and longer onset latency. In most of the subjects this second EMG response could be divided into two parts : a ML (46.2 ± 4.9 ms) and a LL (75.5 ± 8.8 ms) bursts (cf figure 1).

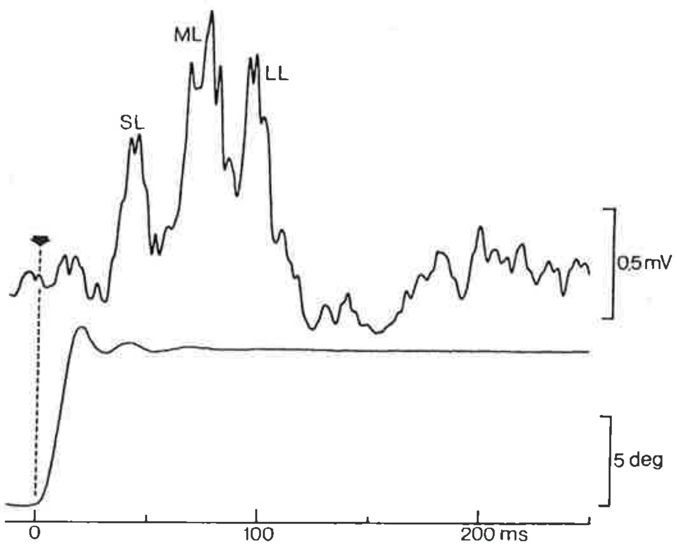

Fig. 1 EMG response of the first dorsal interosseous (FDI) to an imposed displacement of the index. Each record is the average of 32 responses. SL : short latency, ML : medium latency, LL : long latency

During fatigue induced by voluntary contractions and by electrically triggered contractions, the latencies and durations of the different EMG components were not significantly changed. The amplitude of SL and ML components showed an overall reduction in both fatigue tests, whereas LL did not. The comparison of the normalized amplitude of the EMG reflex components in function of the muscle SAP amplitude showed a significant decrease in SL and ML responses after voluntary fatigue. After electrically induced fatigue, SL and ML were not significantly different from control whereas LL was enhanced. In both fatigue tests, the muscle SAP and the normalized amplitude of the SL and ML responses to the stretch returned to control values within 5 min, whereas the LL component remained enhanced after 15 min recovery following electrically induced fatigue.

In complementary experiments, the electrical stimulation of the skin (below the motor threshold) in the area of the motor point did not change the SL and ML EMG responses to a quick stretch, but significantly increased the LL component. On the other hand, a subcutaneous injection of Lidocaïne in the same area decreased the LL EMG activity by 48 % without significant effect on SL and ML components.

DISCUSSION

Changes in the amplitude of an reflex EMG activity can result from modifications of the excitability of the reflex loop and/or of the synchronization of the muscle fibres action potentials. In these experiments the second possibility should be excluded because the latencies and the durations of the reflex EMG components were not significantly different in control and fatigued muscles. In order to approach the effects of fatigue on the sensitivity of the reflex loop more specifically, the reflex EMG responses have been normalized in function of the muscle SAP evoked by the supramaximal stimulation of the motor nerve.

In voluntary fatigue, the normalized LL is not different from control but ML is significantly decreased. In electrically evoked fatigue, ML is not different from control but LL is significantly enhanced. These different behaviours of ML and LL components during our fatigue tests suggests that they are different entities of the longer latency EMG complex. This point is also supported by the observation that ML returns to control values in about 5 min while LL is still above control 15 min after electrically induced fatigue. The different behaviour of ML and LL during both fatigue tests could be explained by afferents on the MNs originating from different receptors. This point of discussion is coherent with the observations that the electrical stimulation of the digital nerves (10) and the activation of tactile receptors (11) evoke long latency reflex components. Matthews (3) also provided experimental evidence that fast cutaneous afferents contribute to the long latency stretch reflex activities.

When the skin was electrically stimulated at the belly of the muscle and below the motor threshold, we observed a long lasting (15 min) enhancement of the LL response without any significant change in SL and ML. A long lasting excitability change in cutaneous afferents was recently reported by Applegate and Burke (12). Thus, as compared to SL and ML, LL would be more specifically controlled by afferents originated in cutaneous receptors. This point of view is supported by our finding that the subcutaneous injection of 2 ml 1 % Lidocaïne drastically reduces LL and has very little effect on SL and ML.

It is concluded from the different behaviour of ML and LL during fatigue that in the FDI the long latency reflex EMG complex should not be considered as a whole. It is proposed that LL is controlled at least in part by afferents generated by cutaneous receptors.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

NEURAL COMPENSATION FOR MUSCLE FATIGUE IN MAN

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INTRODUCTION

Reflex pathways arising from muscle receptors have long been of interest because these simple reflex arcs have the capability to significantly reduce the burden on the central nervous system, which must control movement in the face of complex, time-varying muscle properties such as fatigue. Muscle force is generally considered to be the output quantity of muscle; the Golgi tendon organ is known to be a sensitive transducer of muscle force (3) and to have the proper reflex connections to close a force-regulating feedback pathway (1). Although most estimates of the strength of force feedback have been too low to confirm an important role for this pathway in normal control of movement (6), the majority of these studies were performed on the decerebrate cat, where the tonic inhibition of tendon organ reflex pathways by descending systems (2) leaves the significance of these results in doubt.

In an earlier study (5), it was found that force regulation provided significant compensation for fatigue-induced muscle weakness in the elbow flexor muscles of normal human subjects. The work described below again employed fatigue as a disturbance of the neuromuscular system, but the use of unpredictable stochastic positional perturbations minimized distortion due to voluntary reactions and allowed the mechanical properties of the joint to be determined both before and after fatigue.

METHODS

A DC motor, operating as a position servo, was used to apply stochastic perturbations to the elbow joints of 8 subjects. A fiberglass cast applied to the forearm of each subject was firmly secured to an aluminum arm-rest bolted to the motor shaft. Subjects were seated in a chair next to the motor such that the humerus was horizontal and the elbow rotated about the same axis as the motor. A bandlimited Gaussian white noise sequence (amplitude ±6°, bandwidth 15 Hz) was applied to the actuator through the computer D/A converter. The first second of data collection was under isometric conditions to allow evaluation of isometric torque-EMG characteristics.

Torque and angle signals were recorded from transducers built into the device. Electromyograms (EMGs) of the biceps, brachioradialis, and triceps muscles were recorded using surface electrodes. The EMG signals were high pass filtered (4.5 Hz), and all of the signals were lowpass filtered at 100 Hz prior to sampling at 250 Hz. An example of the torque, angle, and flexor EMGs recorded in a single trial is shown in Figure 1.

The maximum voluntary flexor torque was determined for each subject, with subsequent trials performed at 7 mean torque levels in the range of 0 - 40% of this maximum. The subjects were asked to maintain a particular mean contraction level while the stochastic angular perturbation was imposed upon the elbow joint. When these trials were completed, fatigue of the elbow flexor muscles was produced by having the subject repeatedly lift a 5 to 10 kg weight with these muscles for approximately 15 minutes. After allowing 10 minutes for the transient metabolic effects of fatigue to subside, an identical set of perturbation trials was then performed.

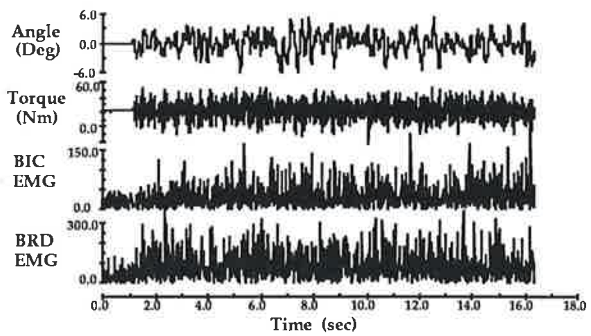

Figure 1. **Example of data collected in typical experimental trial.**

Standard system identification techniques (4) were used to characterize the mechanical state of the elbow joint as compliance impulse responses (CIRs), using torque as the input and angle as the output. These CIRs were found to have the form of a second order mechanical system, the parameters of which (elastic and viscous stiffness, and joint inertia) were used as summary variables of joint behavior.

RESULTS

Isometric torque-EMG relationship and fatigue. The slope of the isometric torque-EMG relationship was used to estimate the impact of the exercise routine on the contractile state of the elbow flexor muscles. This relationship was quite linear (average R^2 of 0.93 across all 8 subjects, linear fit statistically significant in all cases for p=0.95). The average shift in the slope of 247% following fatigue across all subjects means that a given amount of neural drive generated less than one-third of the pre-fatigue torque level, illustrating the enormous deficit inflicted upon the neuromuscular system in these experiments.

Compliance impulse responses and the mechanical impact of fatigue. Compliance impulse responses were identified for each trial (4). Figure 2 (left) gives an example of the general form of the CIR, and illustrates the increase in compliance (decrease in stiffness) and the slowing of its time course following fatigue. A second order mechanical model fit to these responses accounted for 75 - 93% of the output variance, and clearly captured the essential mechanical characteristics of the joint. The elastic and viscous elements of this model are plotted in Figure 2 (right) for a range of mean torque levels, both before and after fatigue (joint

inertia is not shown since it was constant for all conditions). The basic behavior illustrated here, a decrease in elastic stiffness and no change in viscous stiffness, was found in all subjects, with an average decrease in elastic stiffness of 16 Nm/rad and an insignificant increase of 0.091 Nm/rad/sec in viscous stiffness. The change in elastic stiffness is likely the result of a decrease in the magnitude of the reflex component of joint torque due to fatigue-induced muscle weakness.

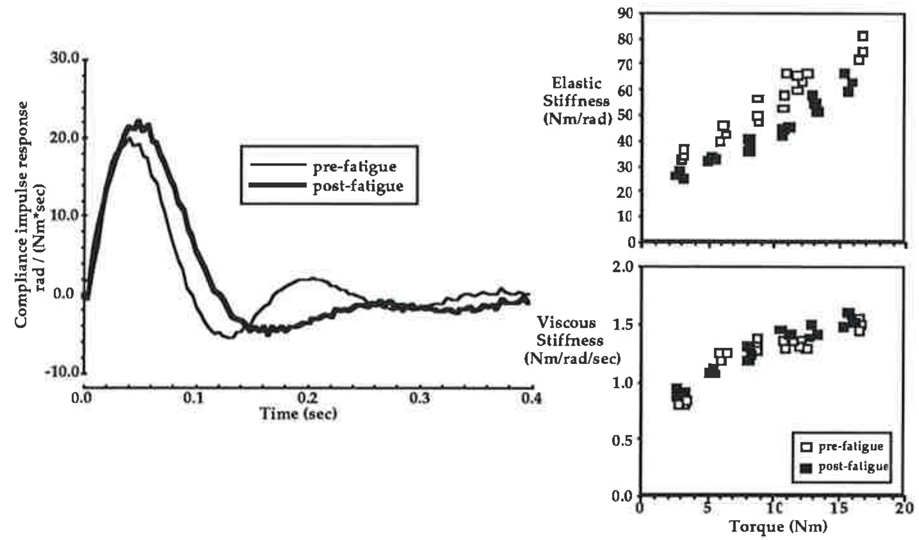

Figure 2. **The effect of fatigue on compliance properties of the elbow joint.**

Effect of fatigue on reflex EMG responses. The average incremental EMG response elicited by the stochastic perturbation was computed for each trial by subtracting the average rectified EMG during the isometric period from the average EMG during the perturbation. An example of the resulting pre- and post-fatigue EMG averages are shown as a function of mean torque in Figure 3. Except for passive trials, the post-fatigue points are consistently larger than the pre-fatigue points, indicating that the reflex component of the EMG response was increased following fatigue. It should bee noted that this increase was above and beyond the operating point shift described by the isometric torque-EMG relationship. The shift in incremental EMG was statistically significant (p=0.95) for all 8 subjects, with an average shift of 161%.

Loop gain estimates. Loop gain is a dimensionless quantity commonly used to quantify the effectiveness of a feedback pathway. A equation for estimating the loop gain of force feedback was derived by comparing the change in elastic stiffness following fatigue to that predicted by a simple model of reflex action, given the degree of contractile loss. For a contraction level of 25% of maximum, average loop gain estimates for all 8 subjects ranged from 0.71 to 3.57, depending on whether reflex action was assumed to account for 30 or 50% of the total joint stiffness. These loop gain estimates translate into a reduction in the sensitivity of the overall system to fatigue of 42 - 78%.

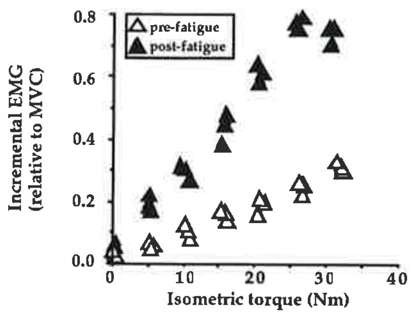

Figure 3. Effect of fatigue on perturbation-evoked EMG responses.

DISCUSSION

It has been shown that the stiffness properties of the human elbow joint change much less following fatigue than would be expected given the degree of muscle weakness, and that this relative preservation of mechanical state is apparently mediated by a large increase in the reflexively evoked EMG responses following fatigue. These results indicate the actions of a moderate gain force feedback pathway; loop gains estimates indicate that the actions of force-sensitive reflex pathways significantly reduce the sensitivity of the neuromuscular system to fatigue. If these loop gain magnitudes are common during non-fatiguing conditions, it appears that force-sensitive reflex pathways contribute significantly to the control of muscle contraction.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

CONCURRENT MEASUREMENT OF MUSCLE FATIGUE BY EMG AND 31P-NMR SPECTROSCOPY

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INTRODUCTION

The compression of the power density spectrum of the electromyogram (EMG) during fatiguing contractions has not been definitively explained in physiological terms. Changes in the EMG spectrum may be the result of several factors that include action potential conduction velocity, muscle unit recruitment and ionic gradients at the sarcolemma [De Luca, 1985)]. Our understanding of this relationship is presently based on theoretical conjecture and indirect empirical studies. Further validation of EMG spectral estimates as an objective measure of muscle fatigue must be based in part on demonstrating that changes in the EMG waveform are associated with specific metabolic events that result in muscle fatigue.

Intramuscular lactic acidosis and accumulation of H+ is frequently proposed as a correlate to EMG spectral compression during fatigue [De Luca, 1985]. In support of this hypothesis, H+ ions reduce the excitability of the sarcolemma which in turn decreases the action potential conduction velocity resulting in a compression of its spectra to lower frequencies. More direct evidence that spectral measures of fatigue are related to H+ ions or other metabolites is unavailable because of the difficulty in measuring muscle cell energetics non-invasively during human volitional contractions. Recent developments in phosphorous magnetic resonance spectroscopy (31P-NMR) may provide the first effective means to non-invasively and continuously monitor intramuscular pH and muscle metabolism while normal electrophysiological function is maintained. This paper describes the simultaneous measurement of EMG, torque production and metabolite levels by 31P-NMR. The emphasis of this paper will be on the concurrent *implementation* of these techniques followed by a brief description of results from a single experiment on the tibialis anterior muscle.

METHODS

The major components of a combined EMG, NMR and torque measurement system are displayed in Figure 1. The EMG and NMR probe are placed over the muscle of interest. A torque measuring device for the ankle is shown with torque-feedback provided by an LED display which is viewed using a mirror. Features of these components and their ability to concurrently measure of EMG, NMR and torque without artifact are further described.

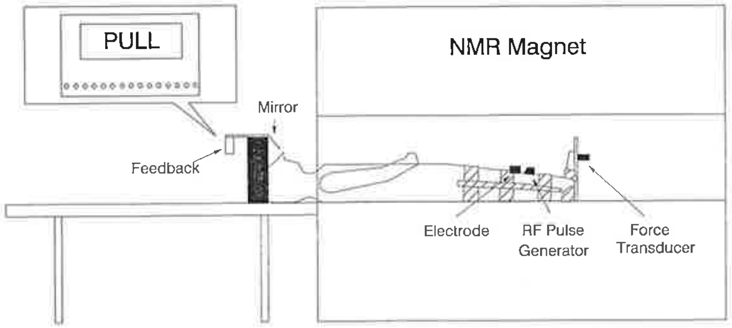

Fig. 1: Schematic of combined EMG and NMR experiment.

EMG Electrode: Surface 31P-NMR spectroscopy necessitates placing the muscle being studied in a large static magnetic field (1-2 Tesla) and then pulsing the NMR probe at a frequency specific to phosphorous to produce a signal that can be analyzed to identify phosphorylated compounds in the muscle (figure 2). The primary considerations for electrode design are that the electrode circuitry a) must not interfere with the homogeneity of the static magnet field and b) must not be contaminated by the RF signals of the NMR probe. Active, rather than passive electrode design was considered because of the need to reduce artifact attributable to capacitive coupling from the lengthy electrode cables (8 ft) needed to reach the center of the magnet's core. To prevent the RF pulses from entering the non-ferrous electronic circuit of the electrode, the circuitry and leads were housed in two layers of copper Faraday shielding with all inputs and outputs lowpass filtered. Artifact was effectively reduced in the EMG signal from a magnitude of approximately 1 mV to a final value of 20 uV with a duration of only 1 ms. The source of this artifact was assumed to be partially attributed to demodulation of the RF signal by the non-linear elements of the operational amplifiers comprising the electrode circuit. Attempts at reducing the artifact to an acceptable level when the EMG electrode was centered directly beneath the NMR probe were unsuccessful. More effective shielding and

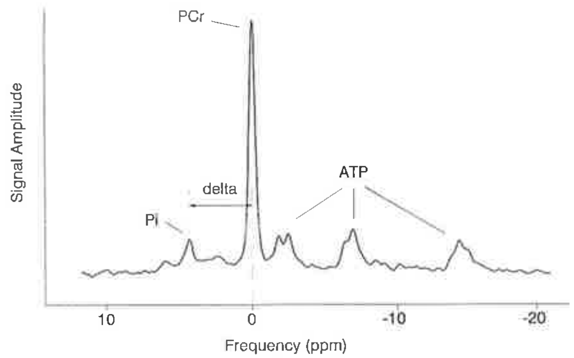

Fig. 2: 31P-NMR spectra for tibialis anterior (at rest).

gating the EMG with the RF pulse are being investigated as possible solutions.

In addition to locating the EMG and NMR probes as close to each other as possible, the volumetric area of detection should also be matched by specifying appropriate coil sizes and shapes to match interelectrode distance and geometry of the EMG probe. A compromise must be achieved to maximize the coil size to increase the inherently weak NMR signal while also recognizing that similar increases in interelectrode seperation increase the likelihood of "crosstalk" from neighboring muscles. Exciting prospects for enhancing spatial resolution are being developed using magnetic field gradient techniques [Sapega et al, 1987].

Muscle Torque: Constant-force, isometric contractions are controlled by measurement of the torque exerted by the muscle being studied. For experiments on the tibialis anterior muscle, two low-compliance load cell transducers are positioned to measure torque about the ankle. The signals from these load cells are directed as feedback to an LED torque display which is visible to the subject via a mirror. The display indicates the desired percent of maximum voluntary contraction (%MVC) as a target level to be maintained during each contraction. The display and an additional timer circuit indicate to the subject when to perform each contraction.

PROTOCOL

The RF pulse sequences needed to improve the S/N of the NMR signal limits its time resolution compared to the EMG technique. Therefore, exercise protocols to produce fatigue in the muscle must specify target force levels, sustained durations or duty cycles that are appropriate to allow a sufficient number of NMR data points to characterize the behavior of the muscle metabolites. As an example, data is presented from one subject for a series of duty cycle contractions (6s on, 3s off) for a duration of 9 minutes followed by 6 minutes of recovery. Contractile force levels were set at 60 %MVC for the first 4.5 minutes of exercise followed by 50 %MVC for the remaining 4.5 minutes of exercise. This particular exercise protocol was designed to induce a significant reduction in intracellular pH at a gradual rate which was then maintained at a steady-state level until the recovery period.

RESULTS

NMR spectra from each of the phases of the exercise are presented in figure 3. The 60 %MVC contraction resulted in a noticeable increase in the Pi peak with a concurrent reduction in the PCr peak. These changes were sustained during the 50 %MVC contractions and then returned to baseline during recovery. The behavior of the median frequency parameter of the EMG and the intracellular pH are plotted as a function of the duration of the exercise protocol (figure 4). Both measures were normalized with respect to the baseline values and the

EMG data is smoothed using a lowess procedure. The median frequency appears to decrease more rapidly than the pH. This result is consistent with previously reported studies comparing median frequency and conduction velocity of the EMG. The very different timing and rates of recovery between pH and median frequency is further evidence that H+ ions accumulation at the sarcolemma cannot fully explain the spectral compression of the EMG during fatigue.

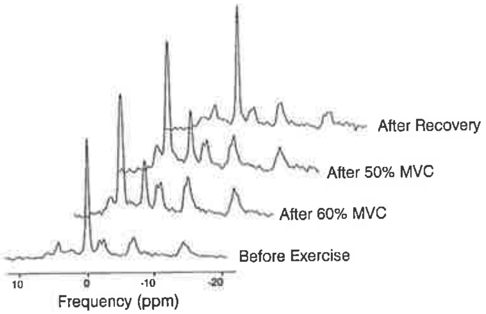

Fig. 3: 31P-NMR spectra for exercise protocol on tibialis anterior muscle.

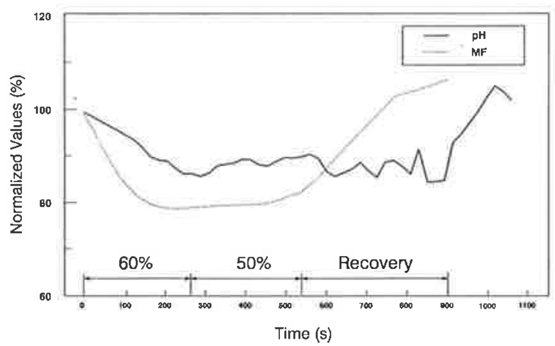

Fig. 4: Concurrent normalized measurement of EMG median frequency and intracellular pH in the tibialis anterior.

ACKNOWLEDGEMENT

This research was supported by the Liberty Mutual Insurance Company and the Department of Radiology, Brigham and Women's Hospital, Boston, Massachusetts.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

USE OF EMG SPECTRAL PARAMETERS TO EVALUATE FATIGUE ASSOCIATED WITH PRESSURE GLOVE WORK

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INTRODUCTION

Fatigue of the hand and forearm during Extravehicular Activities (EVA) is an important factor in determining human work productivity during manned space shuttle operations. Astronauts have reported that hand fatigue during pressurized glove work is a limiting factor in EVA productivity [O'Hara et al., 1988]. Few studies have assessed the effects of EVA gloves on basic hand capabilities. Furthermore, no studies have provided quantitative physiological measurements of work tasks utilizing EVA gloves.

This report describes an assessment procedure of muscle fatigue during a simulated EVA work task. Because fatigue is often defined as a multidimensional process, both physiological and subjective measures of fatigue were included. Physiological fatigue was assessed by recording the median frequency of the electromyographic (EMG) signal from two muscles of the forearm. The specific goals of this investigation were to demonstrate the feasability of using EMG median frequency estimates to evaluate muscle fatigue associated with the use of EVA gloves and to delineate the relative effects of the EVA glove and differential pressure on the fatigue indices. Together, these objectives could provide a standardized test methodology for use by engineers and researchers in the assessment of human factors related to EVA glove design.

METHODS

Ten male subjects (age range: 22-55 years) utilized a glove box to test barehand, gloved hand-0 psid and gloved hand-4.3 psid during a fatigue inducing gripping task. For this study, the NASA Series 1000 pressure glove (operating pressure, 4.3 psid) was used. Two muscles of the forearm were selected for EMG monitoring: the flexor digitorium superficialis (FDS), a finger flexor and the extensor carpi ulnaris (ECU), a wrist extensor. Specially constructed low-profile (<1mm), active surface EMG electrodes were adhered to the muscles described. All EMG signals were recorded on a multichannel instrumentation recorder for later analysis.

The test protocol began with a determination of the subject's maximum voluntary contraction (MVC) by squeezing a hydraulic hand dynamometer in both the barehand and gloved hand (4.3 psid) condition. The barehand MVC value was used to standardize further test contractions and normalize the EMG so comparisons could be made across subjects. Subjects then sustained a 10 s constant-force isometric contraction at 20% of barehand MVC using the hand dynamometer. EMG was recorded during this 10 s test contraction. Immediately after completing this contraction, subjects began a "dynamic" fatigue sequence using a gripping fixture similar to a bicycle brake handle that was coupled to a Baltimore Thereapeutic Equipment (BTE) Work Simulator . Subjects were required to squeeze the BTE gripping fixture at a constant rate of 45 contractions per minute for a total of one minute. EMG was not recorded during this 1 minute of dynamic contractions. The amount of work performed was calculated by the BTE Work Simulator for each cycle. This sequence of a brief static contraction at 20 %MVC followed by 1 minute of dynamic contractions using the gripping fixture was repeated for a total of six times . Immediately following these six fatigue sequences, a "recovery" sequence was initiated in which a series of 20 %MVC X 5s contractions were produced following either 30 s or 2 min rest periods. EMG was recorded only during the 20 %MVC contractions and not during the rest periods. At the end of each isometric contraction the subject gave a subjective fatigue rating on a five-point scale where a value of 5 corresponded to complete fatigue.

The EMG signals were processed with a device called a Muscle Fatigue Monitor (MFM) to compute the median frequency of the EMG signal [Gilmore and De Luca, 1985]. The initial median frequency (IMF) was obtained seperately for each isometric contraction by calculating the y-intercept of a straight line regression fit by a least-squares method to the median frequency data. Fatigue resulting from the gripping task was therefore quantified by the decrease in the IMF values as a function of the different test trials.

RESULTS AND DISCUSSION

The average value of the change in IMF for the two muscle groups tested are plotted in Figures 1a and 1b. The FDS resulted in nearly twice as much overall change in IMF than did the ECU. The ECU however demonstrated a measurable decrease in IMF for all three conditions tested whereas the FDS fatigued only during the glove conditions. A comparison of the fatigue curves for these data indicate a nearly opposite pattern of results for the FDS and ECU muscles. For the ECU, the barehand condition exhibited the greatest fatigue and the pressure glove condition the least. When fatigue was present, most of the change in IMF occurred between trial 0 and trial 3 at which time the change in IMF reached a plateau until recovery commenced at trial 7. These results demonstrate that fatigue is dependent on the glove condition as well as the muscle being assessed. The fatigue effects of the pressurized glove can be nearly equally related to the effects of pressure and the glove itself. Likely explanations for these effects are that they alter the stiffness, mobility and sensitivity of the hand.

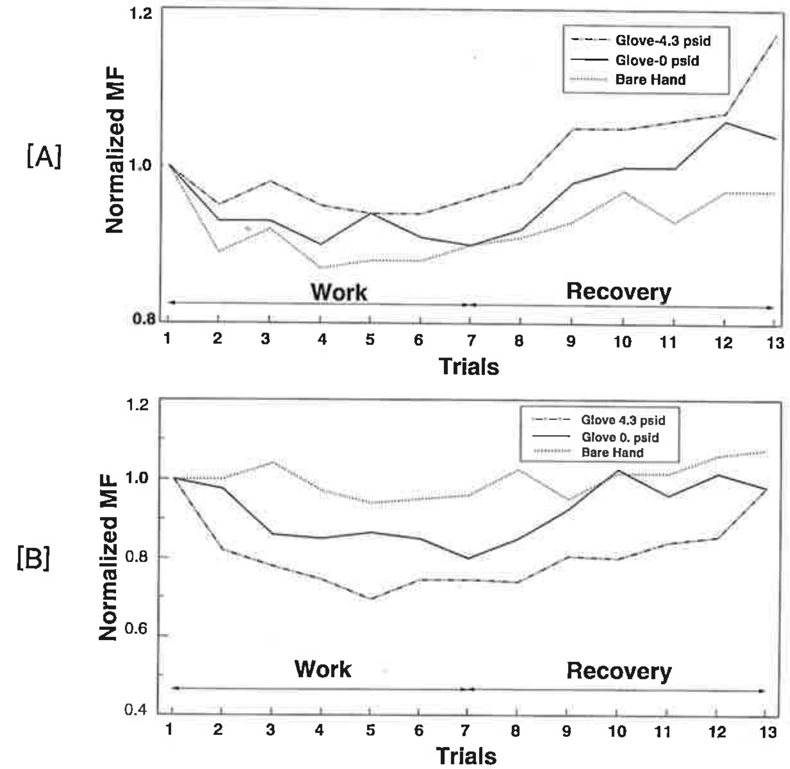

Fig. 1a) Average normalized IMF of the flexor digitorum superficialis muscle for three test conditions; b) Average normalized IMF for the extensor carpi ulnaris muscle for three test conditions.

The average subjective fatigue rating collected during the fatigue protocol are presented in graphic form in Figure 2. While the ratings at Trial 0 were approximately equal for all three glove conditions, the increase in slope is greatest for the glove-4.3 psid condition and proportionately less for the glove-0 psid and barehand conditions. The average fatigue rating (+/- S.D.) for the pressurized glove at trial 6 was 4.42 (0.71) which corresponds to near-exhaustion. The results for the subjective fatigue rating were consistent with the EMG results for the FDS muscle except under the barehand condition where the

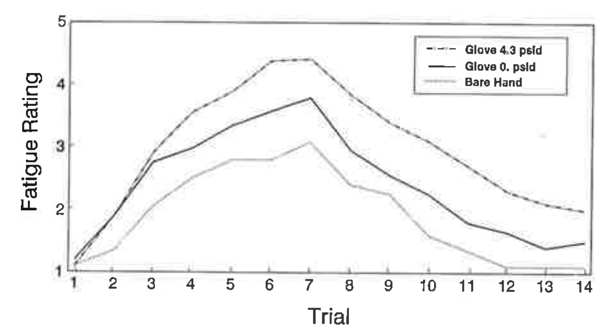

Fig. 2 Average subjective fatigue rating for test conditions.

subjective fatigue rating was relatively greater than the EMG measure of fatigue.

The average amount of work during repetitive gripping over all six fatigue trials was 61% of barehand for the unpressurized glove condition and only 32% of barehand for the pressurized glove condition. The average fatigue work output for the different fatigue trials and glove conditions ranged from 230.98 (120.49) to 54.83 (50.18) in-lb. Performance decay resulting from fatigue effects are presented in Figure 3 where data for trials 2 through 6 were transformed to percents of work produced in trial 1. Data is seperated according to glove condition and a "least-squares" linear regression is fit. Performance decay can therefore be represented as the negative slope of this regression line. Figure 3 indicates a strong effect of the glove-4.3 psid condition on performance decay.

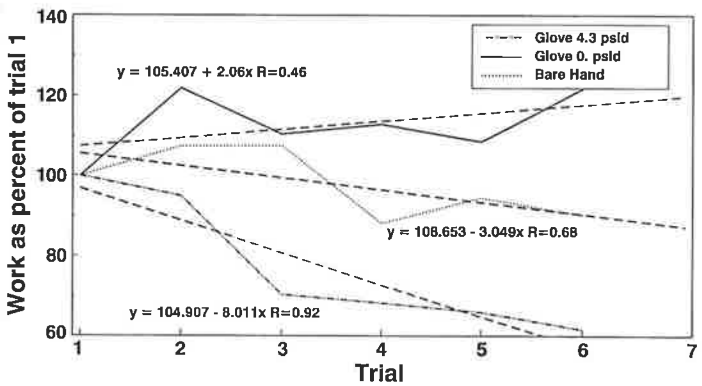

Fig. 3 Average work normalized to trial 1 for the three test conditions.

ACKNOWLEDGEMENT

This research was supported by grants from Grumman Space Systems and Liberty Mutual Insurance Company, Inc. Technical assistance was provided by the International Latex Company, Dover, Delaware.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

A MUSCLE FATIGUE INDEX ASSOCIATED WITH THE EMG EVOKED POTENTIALS DURING FATIGUE

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INTRODUCTION

Conventional fatigue indexes have been mainly discussed in terms of the decrease of the conduction velocity. In this paper, we try to add the information on the muscle fiber type composition to the index by using EMG evoked potential (EEP). The number of the active fast-twitch fiber seems to be more during the EEP than during the background sustained contractions. Moreover, the EEP waveform is a compound action potential. Thus we expect that the influence of the active muscle fiber type composition will appear prominently during the EEP.

As an index we propose the instantaneous frequency of EEP waveform with respect to time. We call this the instantaneous frequency pattern (IFP). The IFP allows us to find every detail of the frequency changes appearing in the EEP. Besides, the IFP showed remarkable changes with the progression of fatigue.

EXPERIMENTAL PROCEDURE

We measured surface EMG signals from masseter muscles. The bipolar surface electrodes were placed on the skin parallel to the muscle fibers and 25 mm apart; the diameter of the electrode was 9 mm. Six male subjects were clenching with an 80% maximal voluntary contraction about a few minutes. The EEP's were measured with a period of 120 cycle/min mechanical chin tapping during the whole exercise.

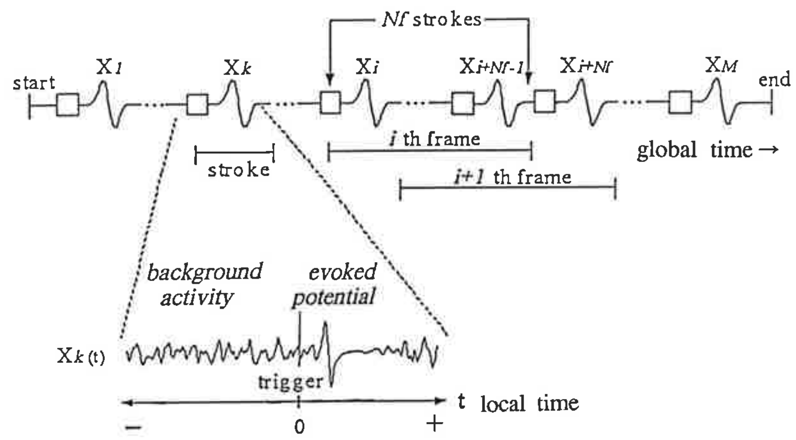

Fig. 1. Pre-processing and estimation for EEP's during fatigue.

After we digitized a signal at a constant sampling rate of 5kHz with a 14 bit analog to digital converter, we extracted the successive EEP waveforms, using the trigger sound of mechanical chin tapping. We pre-processed EEP's with the adaptive correlating filter in order to reduce the influence of the latency jitter. This technique was proposed by Woody (1). That is, estimating the latency jitter between the EEP template and each EEP waveform by the cross-correlation, EEP is appropriately shifted and averaged to yield an improved synchronous averaging.

We defined the local time for each stroke and the global time during the whole exercise (Figure 1). The pre-processing and the estimation were carried out at each frame: one frame contains ten strokes. By sliding the frame along the global time, we estimated the time-course of the mean power frequency (MPF) for background activity and the change of IFP for the EEP waveform.

IFP BY THE ANALYTIC SIGNAL

The basic theory of IFP depends on the recent time-frequency analysis with the analytic signal (2). The derivative of the phase of the analytic signal is the instantaneous frequency. There have been some procedures for forming a complex signal from a real one. Suppressing the negative frequency part and multiplying the positive frequency part by two, we can add to the signal an imaginary part that is the Hilbert transform of the signal.

In my research we defined the analytic signal, $\tilde{x}_i(n)$, by the Hilbert transform of the EEP, $\hat{x}_i(n)$, as follows:

$$\tilde{x}_i(n)=x_i(n)+j\,\hat{x}_i(n) \qquad (1)$$

where $x_i(n)$ is the digitized EEP waveform; n denotes the digitized local time index.
Thus we can estimate the instantaneous frequency

$$f_i(n)=\frac{1}{2\pi}\cdot\theta'(n), \qquad (2)$$

in terms of the instantaneous phase

$$\theta(n)=\tan^{-1}\frac{\hat{x}_i(n)}{x_i(n)}. \qquad (3)$$

RESULTS

Figure 2 shows the typical changes of the EEP waveform and the IFP at each phase of fatigue. We also showed the time-course of MPF as a conventional indicator of fatigue. The broken lines are the references at the frame when the MPF took the highest frequency during non-fatigue phase.

During the early phase of fatigue, the IFP had a mono-peak at the zero-cross point of EEP waveform (zx); a plateau pattern sometimes appeared during the middle phase, besides the significant reduction in the maximum frequency; at the last phase, the IFP showed again a mono-peak pattern, but the peak was changed from the zero-cross point to the first part of the EEP waveform (FST).

DISCUSSION

The IFP is multidimensional at each global time, whereas the conventional ones are one dimensional. Thus we can estimate the degree of fatigue by the IFP without evaluating the whole

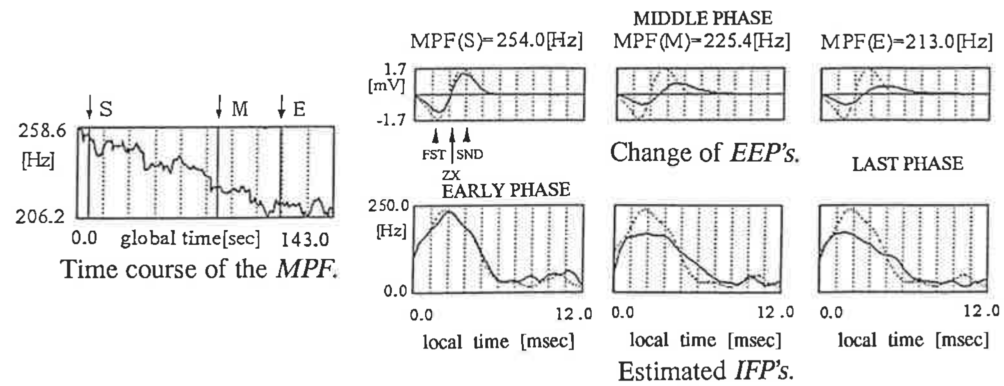

Fig. 2. Analysis of the progression of fatigue. Changes of the EEP waveform and the IFP at each phase (S, M, and E in the global time). Time course of the MPF is showed for reference.

time-course of index during fatigue. However, what caused the remarkable change of IFP has not been cleared yet. We tried to investigate the change of IFP in relation to the fatigue process by using the computer simulation. In Figure 3, first we prolonged the duration twice as long as the raw EEP waveform at the early phase. The result indicated that the IFP's were similar with each other, but the peak of IFP remained at the zero cross point of EEP waveform. This means that the change of IFP was rather complex.

The more detailed simulation (Figure 4) suggested that the fatigue might progress with two different stages regarding the muscle fiber type composition and conduction velocity. At early phase, we had to synthesize the waveform with two sinusoidal waveforms of different frequencies. At middle phase, the plateau IFP was simulated only by one sinusoidal waveform. The prolonging of only the latter half the waveform at the middle phase was needed at the last phase.

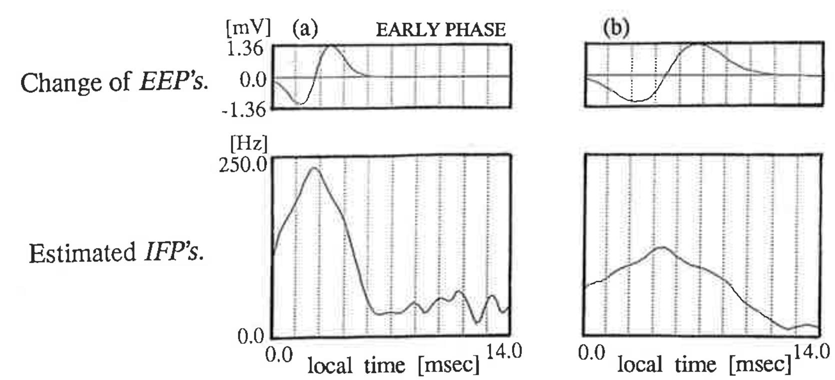

Fig. 3. Examination of the IFP change. (a) The raw EEP waveform at non-fatigue phase.
(b) After we prolonged the duration twice as long as that in (a), the IFP was estimated.

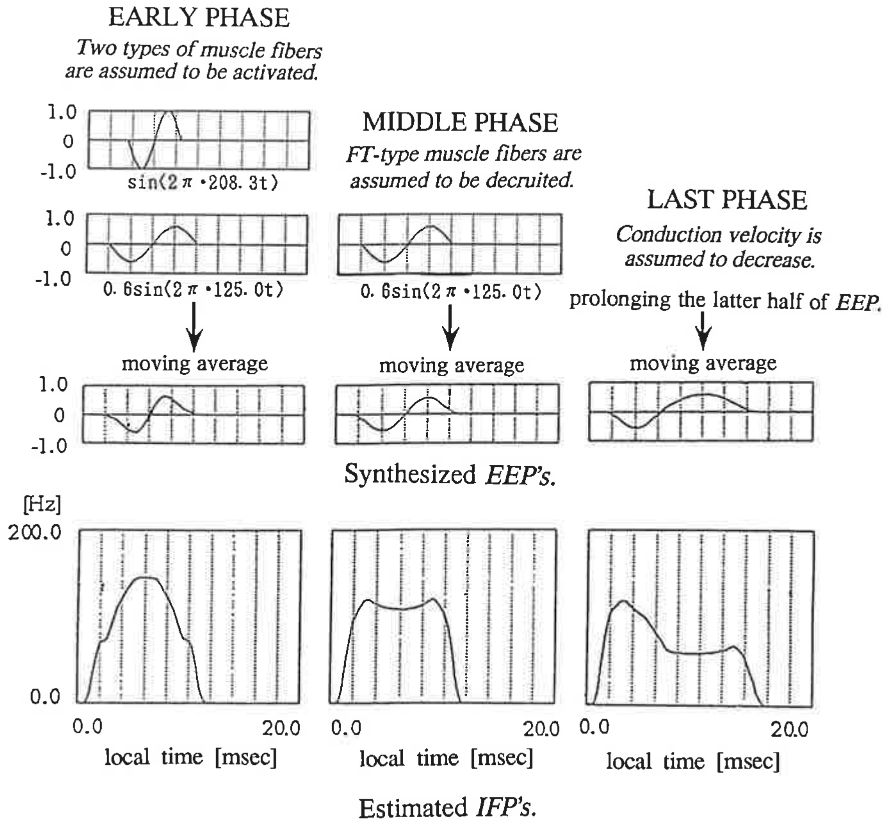

Fig. 4. Examination of the IFP change with computer simulation.

As a result, the dominant factors at middle phase may be something related to the active muscle fiber composition, especially the rate of the active fast twitch fibers. At last phase, the decrease of the conduction velocity probably became prominent.

CONCLUSION

We propose a multidimensional fatigue index by using the instantaneous frequency of EMG evoked potential. The IFP showed the remarkable changes at each phase of fatigue. Thus we can estimate the degree of fatigue by the IFP without evaluating the whole time-course of index during fatigue. This is the different point from the conventional indexes.

We also investigated the mechanism of the IFP change by the computer simulation. The results suggested that the change of IFP may be related to the active muscle fiber composition and the decrease of the conduction velocity.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

COMPRESSIVE STRENGTH OF LUMBAR SPINE ELEMENTS RELATED TO AGE, GENDER, AND OTHER INFLUENCING FACTORS

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INTRODUCTION

In occupational biomechanics simulation models of the human skeleton and musculature have been developed for the analysis of lumbar stress during load manipulation [1, 2], one of the most frequent causes of back disorders and injuries, low back pain and absenteeism [3]. However, it has proven difficult to derive suitable and statistically valid criteria for the assessment of lumbar stress which exclude an increased health risk for persons who manipulate goods professionally [4]. The aim of this study is to quantify the static compressive strength of lumbar spine elements, this being regarded as an adequate measure of the lower spine's maximum load-bearing capacity, and to select and evaluate the most important influencing factors.

METHOD

A number of investigations described in the literature aimed to determine lumbar compressive strength using autopsy material. In the present study, results from measurements on specimens of insufficient size, such as isolated cortex or annulus material, were rejected, as were values from thoracic elements, and segments taken from small children. In some sources the documentation of the conditions obtaining during the investigations was inadequate for the statistical analyses here. A total of 477 strength values from 11 sources [5-15] were considered.

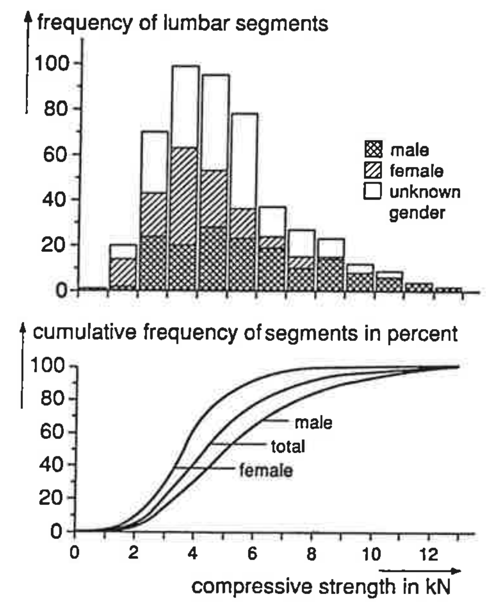

Fig. 1: Frequency and cumulative frequency of gender-related compressive-strength data provided in the literature; number of specimens: 87 [5], 5 [6], 9 [7], 16 [8], 11 [9], 34 [10], 109 [11], 33 [12], 23 [13], 142 [14], 8 [15]

RESULTS

The frequency distribution of the collated strength data is presented in the upper diagram of Fig. 1, and the cumulative frequency curves in the lower diagram. The number of values from male donors amounts to 160, 124 values come from females, and 193 values originate from donors of unknown gender. As Fig. 1 demonstrates, the compressive-strength values are scattered over a wide range. The highest value amounts to 13.0 kN, the lowest being 0.8 kN. The mean value and the standard deviation of all segments amounts to 4.9 ± 2.2 kN. The strength for the male sample is 5.7 ± 2.6 kN, and for the female one 3.9 ± 1.5 kN. The strength quantiles for the samples can be determined from the cumulative frequency distribution. The median in the total sample is approx. 4.4 kN, a strength of between 2 and 6 kN is shown for about 70% of all segments. The median in the male sample amounts to 5.3 kN, and in the female one 3.7 kN.

TABLE I

Influences on compressive strength

	no. of specimens	correlation coefficient
age	167	-0.48
gender	146	0.37
cross-section	270	0.28
lumbar level	441	-0.27
structure	441	-0.20

Table I lists those factors which influence the strength of the lumbar spine. The values provided for the number of the respective specimens vary since not all of the listed variables are documented in each of the sources. If several strength values referred to one person due to the preparation of various segments from a donor's lumbar spine, one value was chosen at random for the age and gender-related evaluation in this paper. Age and gender are donor attributes and, as such, are only indirectly segment characteristics. (Gender was coded "0" for female and "1" for male donors.) By contrast, the characteristics cross-section or the so-called lumbar level refer directly to the specimen. A lumbar level of "0" was assigned to the lowest segment in the lumbar spine (disc L5-S1), the next vertebra L5 received the lumbar-level value "1" etc., with the uppermost lumbar segment, the disc T12-L1, having the value "10". The characteristic "structure" in Table I describes the difference between discs and vertebrae. If the damage during compressive strength measurements occurred in a vertebra, the structure value is "1", it being "0" for disc damage. In addition, Table I indicates the correlation coefficient between lumbar compressive strength and the influencing factors. The negative sign signifies that the strength values decrease with increasing values for age, coded lumbar level and structure. The square of the correlation coeffient is used to quantify the proportion of the variance in strength which is explained by the particular factor and its variation. Accordingly, approx. 23 % of the variance is due to the age difference between the donors and 4% to the segmental structure. The error probability for all variables is less than 0.0001.

The dependence of lumbar compressive strength on age is shown in Fig. 2. The 167 segments in total originate from 167 persons, for 21 of whom the gender was not documented. As Fig. 2 shows, the compressive strength values found in the high age groups are smaller than in the low age groups. Although the strength is greater for male than for female donors, it decreases more rapidly with age. The regression lines indicated in Table II result for linear regression models of the type "strength = a + b • age" if the original

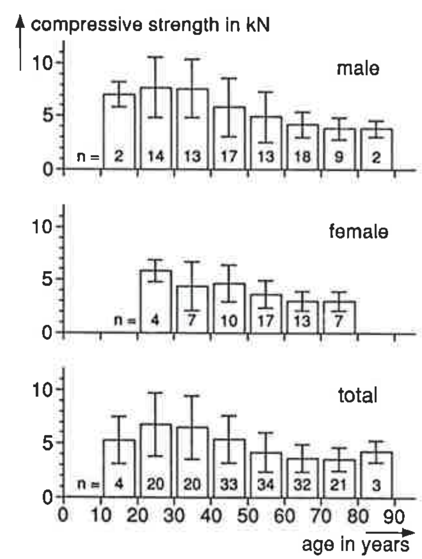

Fig 2: Influence of age on gender-related lumbar compressive strength; n: number of specimens

values for age rather than the classified data in Fig. 2 are assumed.

TABLE II

Linear regression analysis of compressive strength versus age, related to gender

	no. of specimens	intercept [kN]	regression coefficient [kN/decade]	correlation coefficient
male	88	9.87	-0.85	-0.57
female	58	6.78	-0.55	-0.52
total	167	8.23	-0.66	-0.48

The lumbar level represents a further factor influencing compressive strength. Fig. 3 reveals this dependence. The location of the damage during the compression test was assigned for a total of 441 lumbar segments. Within the lumbar spine, caudal elements possess greater strength than cranial segments. The strength is lower for vertebrae than for intervertebral discs. This is due to the fact that, in most cases, the endplate of a vertebra yields and fractures appear before damage occurs in the disc in the form of leakage of the nucleus fluid or rupture of the annulus. The results of regression analyses of the strength according to lumbar level for both the structure-related sub-collectives "vertebrae" and "discs" as well as for the total sample are collated in Table III. According to this, the strength difference between adjacent vertebrae, for example, amounts to 0.44 kN since two units of lumbar level have to be considered.

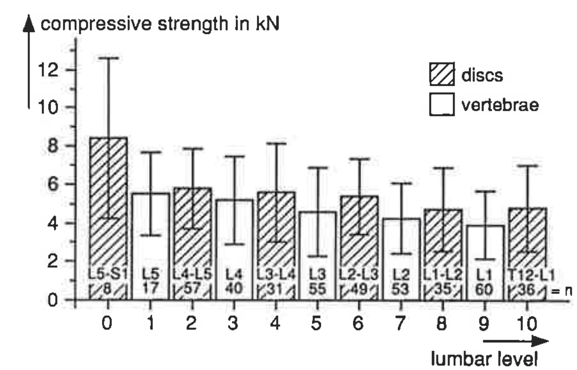

Fig. 3: Influence of lumbar level on compressive strength, related to discs or vertebrae; n: specimen number

The concluding variance analysis is based on the values taken from 111 persons for whom the 5 factors listed in Table I are documented. The following regression equation provides help in predicting the individual strength of a lumbar disc or vertebra from information about characteristics. Only those properties are considered which are readily determinable. Age (A) and gender (G), as well as the lumbar level (L) and structure value (S) for a vertebra or disc, can be determined directly. The cross-section (C) can be established using x-rays or other image-producing methods. If this is not possible, a value of 16.2 cm^2 (mean area of all 270 segments, standard deviation 3.3 cm^2) should be used in approximation. The above mentioned codes apply to gender ("0" female, "1" male), lumbar level ("0" L5-S1, an increment for each lumbar disc or vertebra), and structure ("0" disc, "1" vertebra). The assumed regression model is as follows:

TABLE III

Linear regression analysis of compressive strength versus lumbar level, related to vertebrae or intervertebral discs

	no. of specimens	intercept [kN]	regression coefficient [kN/unit]	correlation coefficient
vertebrae	225	5.78	-0.22	-0.26
discs	216	6.45	-0.19	-0.25
total	441	6.16	-0.21	-0.27

compressive strength / kN = (7.65 + 1.18 G) - (0.502 + 0.382 G) A / decade + (0.035 + 0.127 G) C / cm^2 - 0.167 L / unit - 0.890 S

The regression coefficient is 0.97. According to the model, a predicted strength of 7.0 kN results for the vertebra L2 with an area of 18 cm^2 which was taken from a 30-year-old man, and a strength of 4.5 kN for the disc L3-L4 with an area of 14 cm^2 which originated from a 60-year-old woman.

DISCUSSION

The compressive strength of the lumbar spine is not identical for each person and not homogeneous within a lumbar spine. The wide scattering of the values is mainly due to the influencing factors analysed above. Factors such as profession, diet, illnesses, body weight, or ethnic group, also influence the strength. Furthermore, differences in the test equipment or in the preparation of the specimens between section and measurement may result in different strength values. The influence of both the person-related factors and the measurement conditions are currently being investigated.

Regression equations, representing average values, can be derived from the statistical analyses provided here. When deriving limits for the maximum permissible load on the spine during manual materials handling, lower values should be preferred in order to ensure a sufficient safety margin. Otherwise, an individual's maximum load-bearing capacity might be overestimated. Using the detailed multiple regression function mentioned above in order to determine compressive strength, the result should be reduced, for example, by the standard deviation of the total sample (2.2 kN). If normal distribution was assumed, the lumbar compressive strength among approx. 84 % of persons would be greater than this limit. Another suitable method would be to halve the calculated strength.

In conclusion, lumbar compressive strength varies within a wide range. Age and gender, the factors with the greatest influence, should always be taken into account when estimating an individual's maximum load-bearing capacity. In a detailed determination, the spine's cross-sectional area, the lumbar level, and the structure of the analysed segment should be additionally considered.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

DIFFERENCES BETWEEN MALES AND FEMALES IN EMG AND FATIGUABILITY OF LUMBAR BACK MUSCLES

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INTRODUCTION

The myoelectric signal, as detected with electrodes from the surface of the skin or with indwelling electrodes (EMG), can be regarded as a randomly varying frequency function consisting of a summation of the action potentials of all active motor units, in a certain part of the muscle (Basmaijan & De Luca 1985). Characteristics of this signal, such as amplitude and power spectrum, can be analyzed to assess muscle function in different normal or pathological conditions. Analysis of the frequency content of the myoelectric signal has become an important non-invasive method to provide information regarding the degree of muscle fatigue (e. g. Lindström et al. 1977, Hagberg 1981, Moritani et al. 1982, 1986). The use of surface EMG power spectral analysis has also gained popularity in evaluating the etiology of low back pain (Roy et al. 1989). The hypothesis is that impaired muscle function due to fatigue, could lead to increased load on passive structures, which in turn could cause low back injury (e. g. Seidel et al. 1987). Thus, an early detection of back muscle fatigue could be important to prevent low back pain. When conducting such studies, Roy et al. (1989) have pointed out the importance of categorizing subjects with different backgrounds, regarding factors such as level of fitness and the presence of a low back disorder at the time of testing. The aim of this study was to compare normal men and women with respect to the fatigue characteristics of the frequency and amplitude of the myoelectric signal from the left and right sides of the erector spinae muscles.

METHODS

Ten male and ten female subjects were tested. The mean ages, body heights and masses of the two groups were 29.7±3.7 and 23.8±2.3 years, 1.89±.07 and 1.68±.05 m, 83.2±6.2 and 59.8±5.1 kg for the males and females, respectively. All subjects were habitually active and had a lean body composition. Some had experienced back pain, but none had any periodic or chronic problems with the lower back. Subjects lay prone with the lower body (from the spina iliaca anterior superior) secured to a bench. When the experiment was started, the subjects were instructed to cross the arms in front of the chest and to extend the upper body until the thoracic spine was elevated 25 degrees from the horizontal. The angle was continuously monitored using an inclinometer. This position was to be maintained for 60 seconds. Myoelectrical activity from the erector spinae muscle was detected bilaterally with bipolar silver/silver chloride miniature electrodes (Beckman). The interelectrode distance was kept constant at 10 mm. The electrodes were placed over the most prominent part of the muscle belly of the erector spinae at the L3-level. Myoelectrical signals were differentially preamplified 100 times 5 cm from the electrodes, band-pass filtered between 10 to 1000 Hz (Neurolog NL 125), further amplified and displayed on an oscilloscope for visual inspection. Signals were sampled at 1024 Hz for 1.5 seconds every 6 seconds during the test with an on-line computer system (HP-300), recorded and stored on disc for later off-line analysis.

Myoelectrical signals were analyzed with respect to their amplitude and frequency content using a 512 point Fast Fourier Transform. In this way, ten data points were obtained over 60 s for the root mean square (RMS) value and the mean power frequency (MPF) of the signal (cf. Hino 1977, Moritani & Muro 1987). Statistical differences were tested with Student's two-tailed t-test ($p < 0.05$).

RESULTS

Marked differences in fatiguability were found between males and females. In general, the males could barely maintain the required position for 60s, whereas all females were able to endure the task for at least 90 s. Figure 1 shows changes in MPF and RMS with fatigue for one male and one female subject, respectively. These patterns were present in all subjects with the exception of three males who displayed a more female-like amplitude pattern.

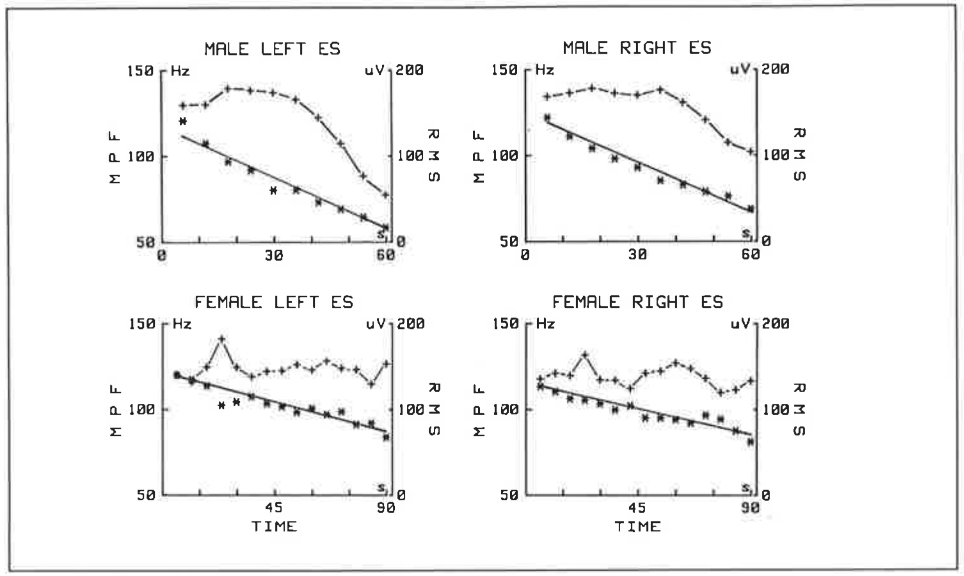

Figure 1. Typical results of myoelectrical signal characteristics from a male (top graphs) and female subject (bottom graphs). Stars indicate mean power frequency (MPF) and plus signs the RMS amplitude of the EMG signal. Note the different time scales for the male and the female subject, respectively.

Mean power frequency (MPF) changes

There was a highly linear decrease in MPF of the myoelectric signal from the erector spinae muscles during 60s of sustained contraction for both males and females (Fig. 2). The linear regression coefficient ranged 0.91-0.99 and 0.93-0.99 for the left and right side in the male subjects, versus 0.87-0.99 and 0.82- 0.99 for the female subjects, respectively. There were no differences between males and females or between left and right erector spinae muscles in this respect. Rate of decline in MPF, however, showed interesting differences between males and females (Fig. 2). Male subjects had a mean decrease of 0.78±0.25 Hz/s and 0.83±0.28 Hz/s for the left and right sides, whereas the corresponding values for the females were 0.54±0.18 Hz/s and 0.48±0.11 Hz/s, respectively. Differences were significant ($p<0.05$). Initial values of the MPF were similar for males and females, as were the MPF's of the left and right sides of erector spinae (Fig. 2).

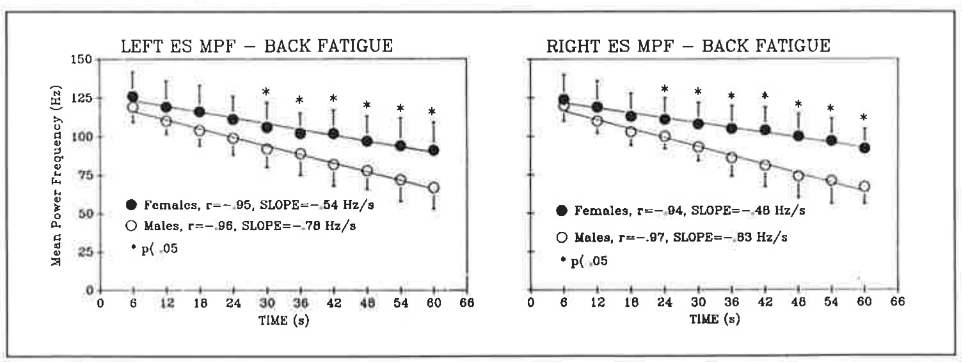

Figure 2. Mean values for the mean power frequency in left and right erector spinae muscles in males and females during fatigue. The lines shown are the lines of linear regression. Asterisks indicate significant differences between males and females ($p<0.05$).

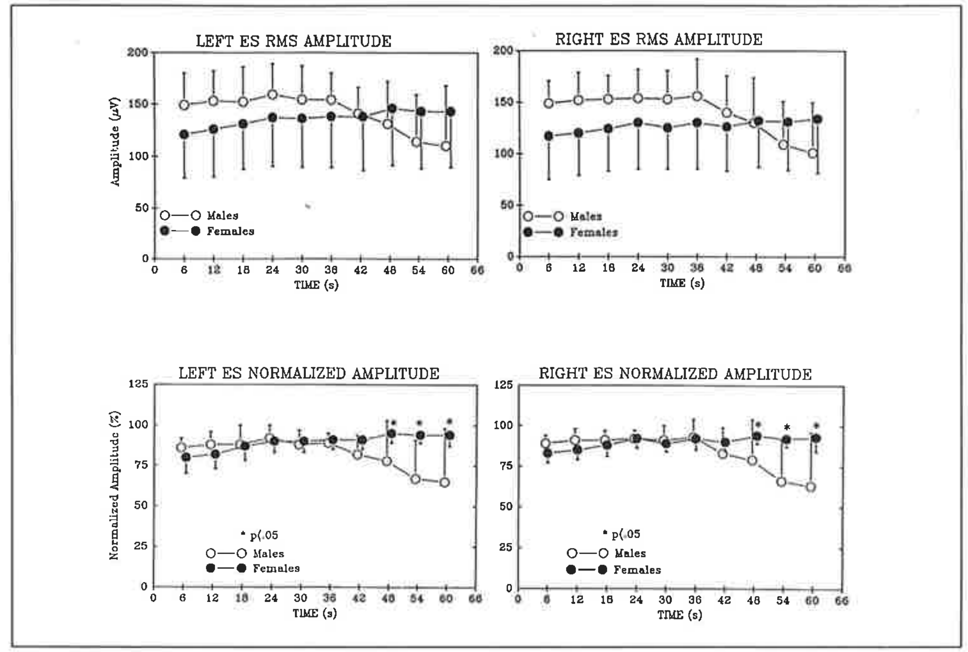

Figure 3. Mean values of the RMS amplitudes (±1SD) for the left and right erector spinae muscles in males and females during fatigue (top). The corresponding amplitudes normalized to the highest RMS value for each individual are shown at the bottom.

EMG amplitude (RMS) changes

Mean values of RMS amplitudes for all male and female subjects are shown in Figure 3. Top graphs show the absolute RMS values for the left and right erector spinae, respectively, with the corresponding normalized values at the bottom. Here the amplitude is expressed in relation to the highest RMS amplitude recorded for each individual during the test.

A dramatic difference between male and female subjects, regarding the changes in the RMS value over 60s of muscle contraction, can be seen. Females showed a slow linear increase in EMG amplitude from 121±42 μV to 142±54 μV for the left side and from 117±42μV to 134±53μV for the right side, respectively. Male subjects, on the other hand, maintained, or increased slightly, the initial EMG amplitude up to approximately 20-30s of the task (from 149±32μV to 153±30μV and 149±22 μV to 156±36μV for the left and right sides, respectively) after which there was a marked drop in amplitude (down to 110±57μV and 102±50 μV for the left and right sides, respectively) (Fig. 3).Toward the end of the 60s contraction, the variation in the normalized EMG amplitude for the males was quite large. This was a result of the fact that 3 subjects maintained the EMG amplitude through the whole test (Fig. 3).

DISCUSSION

The results of this study revealed interesting sex differences in fatiguability of the muscles of the lower back during a sustained isometric contraction. The females were clearly able to better withstand the torque exerted by their upper body. All females endured the task for at least 90 s, whereas most males had problems to maintain the position for 60 s. Analysis of the myoelectric signal revealed sex differences in both the slope of the MPF-time relationship and in the fatigue pattern of the amplitude of the EMG. The males had a decline in MPF with fatigue which was 45% higher for the left and as much as 73% higher for the right erector spinae as compared to the females.

These differences can probably be attributed to several factors. One is that the task of keeping the trunk in an extended position may not have been a standardized load, neither in relation to the maximum voluntary contraction (MVC) of the different subjects nor in relation to dimensional differences between the sexes. An estimation from available anthropometric data, and considering equal force per cross-sectional area of the trunk extensor muscles in males and females (Thorstensson & Oddsson 1982), indicates a 15% higher relative load for the males in this task. Another related factor could be differences in flexibility. A more flexible spine in females would imply a lower relative load on the muscles in the position used.

Also differences between males and females with respect to fibre type composition of back muscles support the findings of this study. Thorstensson & Carlson (1987) reported that lumbar back muscles of females had smaller type II fibres, resulting in as much as a twofold higher type I/type II area ratio than in males. Thus, the relative cross-sectional area occupied by the fatigue resistant type I fibres was larger in females, even though the relative number of type I fibres was the same in both sexes. Considering differences in metabolic profiles between type I and type II fibres, these results suggest a better endurance of female low back muscles, everything else equal. Similar sex differences in fibre size have also been reported for limb muscles (Hedberg & Jansson 1976, Simoneau et al. 1985). It has been suggested that these differences may be attributed to different usage of the muscles in daily activities (Nygaard 1981, Brooke & Engel 1969). Another interpretation of the physiological and functional sex differences seen in this study, is that an increased endurance in lower back muscles in females is beneficial during pregnancy, when the spine is subjected to excessive load for an extended period of time. In this case, the difference could be genetically programmed.

An important consequence of these results is that male and female subjects should be treated as different categories in future studies regarding the assessment of low back muscle function.

ACKNOWLEDGEMENTS

This study was supported by grants from Karolinska institutet, the Swedish Medical Research Council (project no 6529) and the Swedish Work Environment Fund (project no 82-0184). Dr T. Moritani was the first recipient of the scholarship for international exchange of researchers supported by the Swedish Central Association for the Promotion of Sports.

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EFFECTS OF MOTOR UNIT SYNCHRONIZATION AND MEAN FIRING RATE CHANGES ON MYOELECTRIC SIGNAL FREQUENCY SPECTRUM

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Introduction

This report describes a series of experiments that examined the relationship between various motor unit (MU) properties and the spectral changes that occur during fatiguing contractions. Specifically, it will concentrate on two specific parameters: 1) the mean firing rate of the active MUs and 2) the amount of synchronization between the active MUs during constant force contractions.

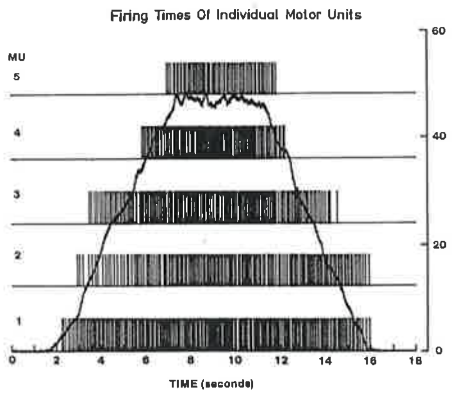
Background

Many researchers have reported on the frequency shift (towards the lower frequencies) of the surface myoelectric (ME) signal during a sustained contraction (see review by De Luca, 1984). Much of this shift has been attributed the decrease in conduction velocity (CV) of the muscle fibres during fatiguing contractions. However, of importance to this paper are the differences reported between spectral parameters and CV decline. Bigland-Ritchie et al. (1981) and Naeije and Zorn (1982) compared ME signal spectral changes with changes in CV. They concluded that changes in CV alone could not account for the simultaneous large changes in the power spectrum. Broman et al. (1985) showed that spectral shifts declined approximately twice as much as CV. Their final conclusion also was that there are factors other than CV that affect the ME signal spectral shift. In particular, they proposed that MU firing pattern changes may significantly contribute to changes observed in ME spectral parameters, such as: 1) a decrease in mean firing rate, 2) a change in the variability of the interfiring intervals of the individual MUs, 3)an increased tendency of grouped firings (i.e. synchronization), and 4) the possible recruitment and/or derecruitment of already active units. These studies suggest the need to examine more closely changes due to fatigue at the MU level, particularly MU firing patterns, in order to explain the different rates of decline in spectral parameters and CV.

Methods

Motor unit action potential trains (MUAPTs) were obtained using a quadrifilar indwelling (needle) electrode. The composite ME signal was decomposed by a proven reliable computer algorithm (see Mambrito and De Luca,1984). A typical firing time plot is shown in Figure 1. Spectral information was obtained from the surface detected ME signal and the needle cannula ME signal. Spectral shifts were monitoring by tracking the median frequency of the spectrum throughout the contraction.

Figure 1: Typical Firing Time Plot for Five Motor Units



Fourteen young, healthy subjects (13 male and 1 female) were recruited for this study. Each subject was seated in a modified dental chair with his/her right foot placed in a restraining-force measurement device. This device allowed for the measurement of ankle dorsi-flexion torque and therefore a measure of tibialis anterior (TA) force. After the subject was seated and his/her foot was positioned in the device, the subject's maximal voluntary contraction was determined. The ME signal was collected from the TA while the subject performed constant force contractions at 80% of their maximal level.

The time-varying mean firing rate of each MU in this study was estimated by passing an impulse train corresponding to the MU firing times through a 400 ms, unit-area, symmetric Hanning window digital filter (similar to technique used by De Luca et al., 1982). The mean firing rate was calculated every two seconds and tabulated. Percent changes in mean firing rate, along with changes in the actual force output of the muscle, were documented for each decomposed motor unit.

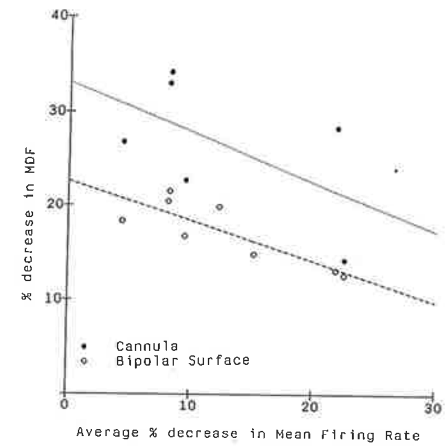
Synchronous behavior of the detected MUs was determined by the construction of cross-interval histograms from the firing time histories. These histograms describe the distribution of first order (i.e. for each MU firing, the nearest forward and backward firing of the other MU in the examined pair is found) firing latencies between a particular motor unit pair. Motor unit pair firings that were between ± 10 ms of each other were defined as being zero-latency synchronized.

Results & Discussion

To investigate the effects of decrease in mean firing rate on the decline in the median frequency (MDF) of the global ME signal obtained from the needle cannula and the surface electrode, the percent change in MDF was plotted against the percent change in mean firing rate as shown in Figure 2. The percent change in mean firing rate was an average of all active MUs in each contraction for the same time period as the corresponding MDF information. Percent changes in cannula and surface-detected MDF were calculated from the initial and final values obtained from the linear regression of the MDF over the specified time range. Only those contractions with linear

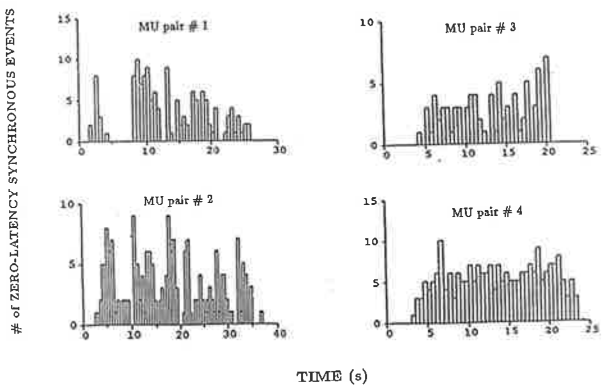
regression correlation coefficients greater than 0.50 were included in the plots to ensure that valid initial and final MDF estimates were used. Contrary to the hypothesis of other researchers, that a decline in mean firing rate may contribute to an increased decline in the MDF of the global ME signal, it appears that the greater percentage decrease in mean firing rate corresponds to the least percentage decrease in MDF of both the cannula and bipolar surface signals. Both plots reveal negative slopes with regression coefficients of 0.56 and 0.83, respectively. These results are contrary to the general model of the ME signal power density spectrum (De Luca, 1984). In this model, the power density spectrum of a MUAP train is a function of the spectrum of the MUAP and the firing rate. Thus, a decrease in firing rate should cause a subsequent decrease in the power spectrum of the ME signal. Our data suggest that changes in MUAP shape decidedly override the effect of decreasing mean firing rate.

Figure 2: Cannula and Surface Median Frequency vs Mean Firing Rate Decline



To examine the role of synchronization on the decline in ME signal MDF, this study concentrated on the amount of synchronization as a function of fatigue by plotting the number of zero-latency synchronous events as a function of contraction time as show in Figure 3. It has been hypothesized that an increase of grouped firings (or synchronization) may lead to additional changes in the ME signal spectrum, unrelated to CV, and thus may contribute to the increased rate of decline in ME signal spectral parameters. Although the incidence of zero-latency synchronization was present in all the contractions analyzed, the occurrences of these events did not consistently increase with a sustained contraction. In fact, in some cases, the number of synchronous events increased and decreased in bursts throughout the contractions. These fluctuations were not observed in the MDF which declined monotonically throughout the contractions. One would expect that if synchronization significantly affects the MDF of the ME signal spectrum that changes in synchronous activity would be reflected by similar changes in MDF.

Figure 3: Examples of Synchronous Events Time Distributions



Conclusion

These experiments showed that mean firing rate decreases were not positively correlated with MDF decreases, but rather the opposite. Thus, MUAP shape changes must have a greater effect than mean firing rate effects. We also saw no consistent trends in the synchronous behaviour of MUs with respect to the changes in ME signal spectra. Therefore, we conclude that these two factors do not play a significant role in the MDF changes that occur during fatiguing contractions.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

USING SURFACE EMG TO STUDY LUMBAR MUSCLE ACTIVITY

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INTRODUCTION

Nearly 90% of all individuals experience an episode of low back pain at some point in their lives [Schuchmann, 1988]. Many cases of low back pain have been associated with abnormal lumbar back muscle activity [Roy et al., 1989]. Thus, an accurate biomechanical model of the lumbar region is a valuable tool in understanding, diagnosing, and prescribing exercises for low back pain patients. In order to validate such a model, a reliable method of evaluating the forces generated in the lumbar muscles was developed. A theoretical concept used to relate muscle force to electromyographic signals is presented which provides a possible validation technique for use with biomechanical models.

METHODOLOGY

A model was developed based on Newtonian mechanics in which the upper body was considered to be a mechanical system in a state of static equilibrium [Ladin et al., 1989]. The model determined the forces generated by the lumbar muscles due to imposed external flexion and lateral bending moments resulting from the position of the upper limbs and any hand-held weights. Data was generated by the model for different bending moment combinations and displayed on the loading plane whose axes are the flexion bending moment and the lateral bending moment. The predicted force of a given muscle was described as a contour map where each curve (termed an isoforce curve) corresponded to loading combinations giving rise to a constant muscular force. Figure 1 is an example of the contour map for the left Erector Spinae Iliocostalis.

Each contour map has three regions of interest. The region where there are no isoforce curves is the inactive region. Any bending moment combinations which result in a point mapped into this region will not produce a force output. The region where there are isoforce curves is the active region for the muscle. The amount of force output produced by the muscle depends on the where the bending moment combinations fall with respect to the isoforce curves. Each isoforce curve represents 30 Newtons of force. The third region of interest is the region where the isoforce curves are very close together. This region is the switching region

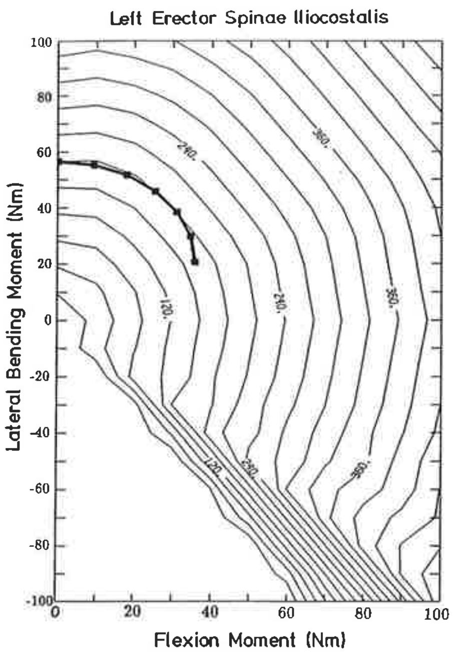

Figure 1: Contour map for the left Erector Spinae Iliocostalis overlayed with bending moment curve obtained from right arm internal rotation exercise

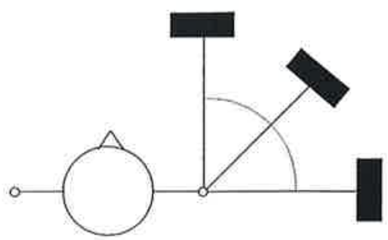

Figure 2: Top down view of right arm internal rotation exercise

and it divides the active region from the inactive region. In the switching region, small changes in the bending moments combinations will produce large changes in the force output.

A simple task was designed which explores the concept of isoforce curves and its relevance to validating the model predictions. A subject held a 4.5 kgf weight in the right hand with the arm horizontally abducted laterally to form a 90^0 angle with the trunk. The subject rotated the arm at a constant velocity by 90^o to the front of the body. Figure 2 displays a top-down view of the internal rotation exercise. The bold curve in Figure 1 is the theoretical set of bending moment combinations resulting from the exercise mapped onto the contour map for the left Iliocostalis. Since the bold curve closely follows an isoforce curve, the left Iliocostalis is predicted by the model to produce a constant force for the duration of the movement.

To validate the model predictions, experiments were conducted using five male subjects who performed the internal rotation task with the right arm holding a 4.5 kgf weight. The RMS value of the amplitude of the electromyographic signal was monitored using a surface electrode on the left Erector Spinae muscle during the exercise. The motion of the arm was monitored using the WATSMART motion analysis system to determine the angle of the arm

with respect to the trunk and the angle that the arm traverses. Planar segments containing infrared light emitting diodes were placed on the elbow of the rotating right arm and on the lateral portion of the trunk beneath the rotating arm. Two WATSMART cameras detected the infrared light from the segments and reconstructed the three-dimensional positions of the segments at every 0.5 second interval. The three-dimensional information was used to calculate the actual bending moment combinations during the exercise.

RESULTS AND DISCUSSION

The left Iliocostalis was predicted to produce a constant force level if the internal rotation moment curves precisely traced the isoforce curves of the contour map. Based on the theoretical internal rotation exercise bending moment combinations shown in Figure 1, the model predicted the muscle would turn on instantaneously at the beginning of each task and maintain a constant force output value for the duration of the task. This ideal was not obtained for all five of the subjects. For example, the contour map of the right Iliocostalis is shown in Figure 3 plotted with the moment curves obtained from subject LG. In the case of LG, the model predictions were slightly different. The muscle was expected to turn on initially and then to continue to increase at a slow rate for the duration of the task. The first plot in Figure 4 displays the muscle force predictions mapped onto the actual EMG signals which were recorded from LG.

For each of the five subjects, the muscle force predictions were calculated based on the intersections of the exercise bending moment curve with the isoforce curves on the contour map. The EMG signals from the five subjects consistently matched the model predictions for at least a portion of the exercise and are displayed in Figure 4. The amplitudes of the EMG signals from JV, JE and KS exhibited a portion during the last two-thirds of the exercise where they were isoforce. LG exhibited a positive slope for the duration of the exercise after the initial rise in activity, and JN exhibited a positive slope for the first half of the exercise and a negative slope for the second half of the exercise. Variations between subjects also existed in the intial slope of the amplitudes of the EMG signals, as well as the maximum amplitude. The maximum amplitude ranged from 0.14 Volts for JV to 0.3 Volts for LG.

CONCLUSION

The introduction of contour maps for the lumbar muscles provided new methods of evaluating and comparing the muscle forces with the EMG signals. The isoforce action of the left Iliocostalis muscle was explored during the right arm internal rotation exercise. EMG signals recorded from the Iliocostalis muscle correlated with the muscle force predictions, although variations between subjects were noted in the initial slopes and maximum EMG amplitudes.

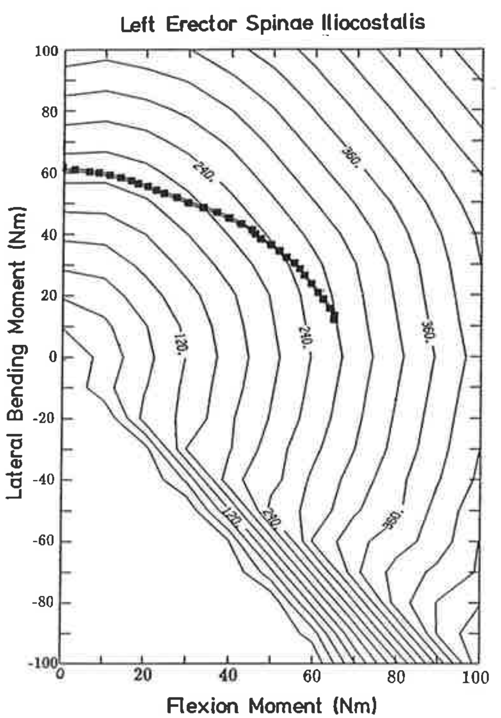

Figure 3: Right arm internal rotation bending moment curve obtained from LG overlayed onto left Erector Spinae Iliocostalis

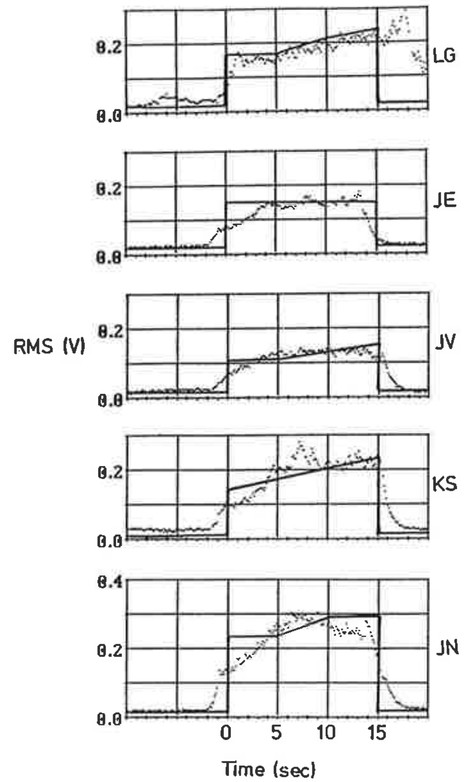

Figure 4: EMG signals and muscle force predictions obtained from all five subjects for the left Erector Spinae Iliocostalis

ACKNOWLEDGEMENT

This research was supported a Patricia Roberts Harris Fellowship.

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Electromyographical Kinesiology
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PROSPECTIVE STUDY OF ISOKINETIC STRENGTH AND INJURY HISTORY FACTORS IN THE PREDICTION OF BACK INJURY

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INTRODUCTION

The nursing profession has a high incidence of low back problems. Studies, conducted mostly in the form of questionnaires, have concluded that both patient transfer and non patient transfer activities contribute to these low back problems. However, normative low back strength data for nurses have not been established. Nor has it been demonstrated that strength, or lack of strength, is an injury predictor in this profession.

The purpose of this study was to test a large group of female nurses and to develop normative values for strength in isokinetic lifting tasks. In addition, the nurses were followed prospectively for two years for any incidence of low back injury. The data were evaluated using discriminant analysis to define predictors of low back injury.

METHODS

The study population consisted of one hundred and seventy-one females employed as nurses in a tertiary care community hospital. Each volunteer was apprised of the purpose of the study, of the risks involved, and gave informed consent. Any volunteer who had back pain at the time of presentation for testing, had ever had back surgery, or had had a back related injury in the six months immediately prior to testing that resulted in absence from work, was excluded from the study and was not tested. Also, no pregnant subjects were tested.

Isokinetic lifting tests were carried out on a prototype linear lift task machine (Cybex, a Division of Lumex, Ronkonkoma, N.Y.). Each subject was instructed as to proper position and lifting technique. The position and technique were demonstrated. The subject then executed five or six supervised warm-up lifts, with the isokinetic mechanism set at a high speed (60 cm/sec). These were utilized as practice lifts so that the correct lifting technique could be observed/instructed, and to provide some muscle warm up. The subject was then asked to exert maximal effort in three test lifts.

The peak force or strength, the total work in the middle third of the height lifted, and the total work done in each lift were averaged over the three lifts for each subject.

Epidemiologic data were also collected via a questionnaire relating to back injury and back pain history over the previous one and six month periods.

Injury reports were monitored to identify any of the subjects who incurred an on-the-job back injury or back pain episode within two years after being enrolled in the study.

RESULTS

There was a significant difference in body weight ($p < .05$) between those nurses who were 30 years of age or younger and those greater than 30 years of age, with the older nurses being 12% heavier. There was also a 15% decrease in strength and work per unit of body weight in the older nurse population.

The individual questions in the questionnaire, as well as factor analyzed groups of questions, were used as independent variables in regression analyses with strength and work used as the dependent variables. The percentage of variance explained in any case was very low, demonstrating that the questionnaire responses were poor predictors of strength and/or work.

The total number of injured nurses in the follow-up period was 16. When the mean peak force generated at the time of the study enrollment for the injured nurses was compared to the non-injured nurses, there was no significant difference. The mean peak force for the injured nurses (N=16) was 60.8 Kg $\pm$ 13.7 Kg and for the non-injured (N=138) 63.7 Kg $\pm$ 13.7 Kg. A Wilks' Discriminant Analysis was carried out to determine the contribution of the lift test and questionnaire data to the total variance of both injured and non-injured groups. None of the questionnaire elements, nor work nor strength entered the equation at a significant level.

DISCUSSION

In an earlier study, Chaffin, et al (1) demonstrated that the statistical incidence of an individual sustaining an on-the-job back injury increased when the job lifting requirements approached or exceeded the individual's strength capabilities. The results of the present study differ from those of Chaffin, et al (1) in that strength did not correlate with the incidence of on-the-job injury. The present study involved only females who generate considerably less maximum force than males (2), the subject population in the Chaffin, et al study (1). Thus, the lifting requirements of a job task may approach or exceed the lifting capacity of a greater portion of females than males regardless of their strength. Also, the on-the-job weights lifted by the subjects of the present study, female nurses in the clinical setting, vary significantly in both magnitude and in the position of the load relative to the body. The Chaffin, et al (1) study was carried out in an industrial setting where the weights lifted did not vary and the loads could be located in ideal positions to facilitate lifting. The variety of loads, the inability to optimally position the load for lifting, and the need to lift and support burdens that can change suddenly in effective weight and shape, result in increased risk and incidence of back injury in the nursing profession that are dominated by factors other than the lifting strength of the individual.

This study raises a question as to what are appropriate predictors for back injury? If strength is not a predictor, what can be used effectively as a predictor? Chaffin, et al (1) present good evidence that the prospects of injury to the low back increase as the job lifting requirements approach or exceed the individual's low back strength. When loads are regulated and do not vary in industry, it appears that strength screening according to the work of Chaffin, et al (1) would be a useful adjunct to the pre-employment history and physical. However, the results of the present study indicate that strength screening may not be effective in reducing injuries in occupations where the loads are not regulated and the conditions of lifting vary considerably and cannot be optimized.

The solution to the problem of back injuries in such occupations would appear to lie not in pre-employment screening, but in the maintenance of physical fitness and in efforts to ensure that the effective loads to be lifted do not exceed the capacity of the lifter. The routes to this solution are numerous but include 1) training/education, 2) redesign of the job, and 3) support personnel or lift assist mechanisms.

ACKNOWLEDGEMENT

This work was supported, in part, by Cybex, a Division of Lumex, Inc., Akron City Hospital, and The Akron City Hospital Foundation.

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Electromyographical Kinesiology
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EMG POWER SPECTRUM OF DISUSED MUSCLE DURING FATIGUE

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INTRODUCTION

It is well known that fatiguing exercises induce changes of the muscle EMG which can be quantified by computing the power density spectrum (PDS). During the course of muscle fatigue, the median frequency (MF), defined as the frequency value which divides the PDS into two part of equal energy content, is shifted towards lower values (1, 2). It has been shown that this shift is mainly related to the reduction in muscle conduction velocity (3, 1).

The aim of this study is to analyse changes of the force and EMG PDS during fatigue, in muscles subjected to a period of immobilization which is known to reduce the action potential amplitude and the maximal firing rate of motor units (MUs) (4).

MATERIAL AND METHODS

Subjects

The effects of disuse atrophy were investigated in the adductor pollicis muscle of seven patients of both sexes immobilized unilaterally for 6 weeks after a forearm fracture and in one subject who volunteered to be immobilized in similar condition. The plaster cast extended from the elbow to the hand and included the thumb. Muscles of immobilized and contralateral hands were compared. Muscle fatigue was studied during a 60s sustained maximal voluntary contraction (MVC).

Force and EMG recordings

The maximal force of contraction was recorded by connecting the first phalanx of the thumb to an isometric strain-gauge transducer (5). The muscle EMG was recorded by using two surface electrodes fixed respectively over the motor point and at the metacarpophalangeal joint of the thumb. The EMG signals were differently amplified (bandwith 6 Hz - 6 KHz) and recorded on magnetic tape by a FM recorder.

EMG processing

Tapes were played back and the EMG signals were digitized at a sampling rate of 5 KHz and low-pass filtered (0-500 Hz) before being stored on floppy disk. The EMG was subjected to power spectral analysis by means of the fast Fourier transformation (FFT) with hanning window from which MF was computed.

RESULTS

In the absence of fatigue, our results show that although the force and the integrated EMG (IEMG) are drastically reduced (- 55 ± 6 % and - 45 ± 7 %, respectively; mean ± SE, n = 8) after immobilization, MF is not statistically different from the contralateral control muscles.

During the sustained MVC, the force and the total PDS decrease progressively whereas MF is shifted towards lower frequencies in both control and immobilized muscles. After 60s of contraction, the maximal force is reduced in similar proportion in control and disused muscles (cf figure 1). On the other hand, EMG parameters are more considerably altered in control conditions. The reduction of the total PDS and the shift of MF toward lower frequencies are larger ($p < 0.05$) in control as compared to immobilized muscles (figure 1).

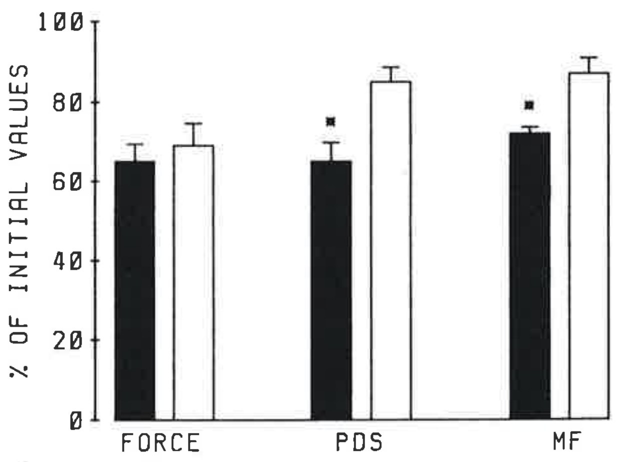

Fig. 1. Histograms showing the effect of fatigue on the reduction of force, total power density spectrum (PDS) and its median frequency (MF) in disuse (light bars) and normal contralateral (dark bars) muscles. Values expressed as percentage of unfatigued muscles are means ± SE, * denotes significant differences ($P < 0.05$)

DISCUSSION

There are different propositions concerning the main cause of the EMG PDS modification. Lindström (3) relates the modification of the spectral shifts to variation of the membrane conduction velocity which subsequently changes the shape of the MU action potential. Other authors suggest that changes in synchronisation of discharges of different MUs and of their firing rate also contribute to PDS modifications (6, 7, 8).

In this work, the force and the IEMG in the absence of fatigue are both drastically reduced after 6 weeks of immobilization. On the other hand, the MF of the PDS is not statistically different from control although the maximal MU firing rate was found to be reduced in disused muscles (4). These last observations comfirm previous work showing little effect of the MUs firing rate change on the MF shift (1). Our finding in normal human muscle of an absence MF change after disuse atrophy is interestingly different from the observation of an MF shift towards higher frequencies in myopathy and towards lower frequencies in neurogenic disorders (9).

It is well known that during sustained MVC, the progressive reduction of force is associated with a slowing of the EMG activity (10, 1). Our observation of a similar reduction of force in both control and disused muscles during fatigue is coherent with previously reported results (11, 12). This work further indicates that the reduction of force is not associated with similar EMG alterations since after immobilization the shift of MF towards lower frequency is smaller than in control muscle. Limited MF shift after disuse could be explained by reduced alteration of membrane ionic processes (cf 13) during fatigue because less high threshold MUs are recruited (14) and their maximal firing rate is smaller (4). This interpretation is coherent with the observation of a larger EMG alteration in disused as compared to control muscles when fatigue is induced by electrical stimulation at a constant 30 Hz frequency (5).

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MOTOR UNIT CONTROL

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

CONTROL OF MOTOR UNIT ACTIVITY: ROLE OF MOTONEURONAL PROPERTIES AND ORGANIZATION

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INTRODUCTION

In this brief summarizing survey, much of my account will be centred on work done within our own laboratory. Most of the experimental results were obtained from hindlimb motoneurones and muscles of cats.

As is well known, the central nervous system controls muscle force by two strategies that are commonly employed in parallel: 1/ by changing the number of active motor units (recruitment gradation) and 2/ by changing the impulse rate of already discharging motoneurones (rate gradation). Below, I will deal with each one of these gradation strategies.

THE HIERARCHY OF ORDERLY RECRUITMENT

As a voluntary or reflex contraction is increasing in strength and progressively more motoneurones become recruited, the various cells are not activited in a random order. Henneman et al. (10) observed the neurones to become recruited in a ranking-order of increasing size, and this behaviour was contained in the summarizing concept of the "size principle" (9). The "size" actually monitored in the original experiments concerned that of the motoneuronal axon (10), and one of the reasons for the functional importance of the ascending-axon-size-order of recruitment lies in the fact that axonal size commonly correlates with essential contractile properties of the muscle units: the thinnest axons tend to innervate units that are relatively slow, weak and fatigue-resistant, and the thickest axons tend to have fast and strong units (1, 9). The ascending-axon-size-order of recruitment represents a case of what one might more generally call property-ranked recruitment, i.e. recruitment ranked in relation to functional unit properties. Such a wider term is needed because, among the relatively high-threshold fast-twitch motoneurones, the recruitment (1, 21) as well as the motoneuronal excitability (e.g. 5, 16) is ranked in relation to muscle unit properties but not clearly in relation to measures of axonal size. However, in spite of these limitations, axonal and other aspects of neuronal size remain important factors in the quantitative and statistical analysis of the organization and behaviour of motoneurone - muscle unit populations.

Already a long time ago, it was found that motoneurones with a slow axonal conduction velocity also had a high input resistance and a low current-threshold for the activation of repetitive impulse firing (12). Among these neurones the differences in current-threshold were probably to an important extent caused by the differences in input resistance; if the various cells had to become depolarized to about the same extent to reach their firing threshold, then less current would obviously be needed for discharging those with a high input resistance than for those with a lower one (cf. Ohm's law). Our continued analysis has indicated that much of this size-associated difference in input resistance depends on differences in membrane resistivity, i.e. on factors that are not causally related to size (17, 18; see also 2, 6, 8, 20). Hence, even in the absence of any specificity in the distribution of active synapses to variously sized neurones (e.g. with an equal excitatory current density for all cells), an ascending-size-order of recruitment would tend to appear. We feel that these results help to explain why such a recruitment pattern is so commonly encountered in motor physiology.

RECRUITMENT GAIN

The ease with which the central nervous system can change the number of discharging cells within a motoneurone-population depends on the intra-pool distribution of recruitment thresholds, as expressed in terms of the intensity of excitatory synaptic current needed for discharging the respective cells. The closer these recruitment thresholds are to each other, the higher becomes the "recruitment gain". The intrinsic differences in neuronal excitability may become enlarged or diminished by suitably distributed synaptic effects. In a recent theoretical analysis we have studied the expected consequences of variously distributed synaptic inputs on recruitment gain (15). We then used a very simple model of a motoneurone pool to demonstrate that not only the synaptic distribution of the command signal itself but also that of any steady "background" activity is of importance. Such background effects on recruitment gain depend on the presence of differences in the intra-pool distribution between these synapses and those of the command signal. Given the appropriate intra-pool distribution, even inhibitory background effects might serve to increase the recruitment gain. Published experimental data on synaptic distributions suggest that synaptic activity might indeed affect the recruitment gain. Thus, for instance, for IA monosynaptic connections from spindle afferents to motoneurones the maximal EPSP-sizes are larger in highly excitable "slow" motoneurones than in the less excitable "fast" ones (1). For skin afferents, on the other hand, the largest excitatory effects were

commonly found in the intrinsically less excitable "fast" motoneurones (3, 7). This latter type of input would be expected to increase the recruitment gain of a pool that was driven by a system with a distribution like that of the IA synapses (15).

TASK-RELATED RECRUITMENT

Even among muscle units that exert their forces in the same direction, different motor tasks may show a preference for different individual units within the same muscle (e.g. 19). Such task-related recruitment patterns may well exist in parallel with a property-ranked strategy such as described above (e.g. "size principle" behaviour). According to the property-ranked strategy the small-axoned and slow units would, statistically speaking, be those most easily activated in the majority of motor tasks. The task-related strategy might then, for instance, determine precisely which ones of the various slow and small-axoned units (e.g. units from different sites; see below) would be those preferentially activated in a given task.

In several cases, task-related differences of recruitment have been found to include a topographical dimension: also within uni-directional (portions of) muscles, different tasks may show a (graded) preference for units located in different muscle regions. This has been very clearly seen in man for the long head of biceps brachii (19), and we have recently been investigating such questions for a uni-directional muscle of the cat's hindlimb, m. peroneus longus (PerL). In anaesthetized animals we found a different antero-posterior distribution of EMG-activity within this muscle depending on whether it was activated by electrical stimulation of the motor cortex or by stimulation of a peripheral skin nerve (flexion reflex) (11). In subsequent experiments on freely moving cats we have found marked and reproducible differences in the antero-posterior distribution of PerL EMG-activity depending on which motor program the animal was executing (Hensbergen and Kernell, in preparation). For the electrophysiological kinesiologist such observations mean that, even for functionally "simple" uni-directional muscles, it might matter considerably precisely where on (or in) a muscle the recording electrodes are positioned; different locations might reveal different activity patterns. For the theoretical interpretation of such differences it is important to realize that there is commonly an evident correlation between the rostro-caudal position of a motoneurone within its spinal pool and the intramuscular site of its nerve endings. For the cat's PerL, rostral portions of the motoneurone pool distribute their endings predominantly to anterior muscle regions and vice versa (4). Thus, for the PerL, the observed antero-posterior differences in EMG-activity probably reflected a rostro-caudal difference in the intra-spinal distribution of motoneuronal activation (for further comments, see ref.14).

RATE GRADATION

As motoneurones become activated during a maintained contraction, their repetitive impulse discharges are evoked by persisting excitatory currents, produced by the summation of many asynchronous postsynaptic events. The discharge-generating action of these postsynaptic currents can be imitated by currents that are injected through the tip of an intracellular microelectrode. Such experiments have demonstrated that the repetitive discharge properties of motoneurones are well matched, in several ways, to the contractile characteristics of their muscle units. The relevant motoneuronal properties include the range of possible discharge rates, the frequency-current relation, and the initial and late phase of frequency-adaptation (for review, see ref.13). Thus, for instance, the minimum rate of maintained firing of a motoneurone is typically such that it corresponds to the lower end of the steep region of its units' tension-frequency relation. Due to this rate-match between motoneurones and their muscle fibers, an increase of discharge rate of a barely recruited motoneurone will indeed lead to a significant enhancement of motor unit force.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

MOTOR UNIT RECRUITMENT ORDER IN VOLUNTARY AND ELECTRICALLY ELICITED CONTRACTIONS

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INTRODUCTION

It is well known that, during voluntary contractions, motor units are recruited according to the ***size principle***. Andreassen [1] has recently demonstrated that muscle fiber conduction velocity (CV) can be considered a "new size principle parameter", a fact confirmed by other authors [5,8,14]. Therefore, the motor unit recruitment order may be reflected by electrophysiological variables such as CV, the median frequency (MDF) or the mean frequency (MNF) of the myoelectric signal power spectrum [3]. It is also known that during direct electrical stimulation of a nerve, the recruitment order is reversed with respect to the ***size principle*** and is controlled by the electrical excitability threshold of the fibers, and by the current density. The recruitment order of motor units with increasing intensity of surface stimulation of a muscle motor point has not been investigated and it is relevant in Functional Electrical Stimulation (FES) applications and in muscle fatigue studies. Since CV and spectral variables are affected by recruitment order, they could be used to provide a global estimate of such order in either voluntary or electrically elicited contractions.

METHODS

To investigate the relationship between surface myoelectric signal variables and the level of voluntary or electrically elicited contractions, the tibialis anterior muscle was studied in two groups of healthy human subjects using monopolar surface stimulation of a muscle motor point, and detection of the single and double-differential myoelectric signals using techniques described in the literature [4,7,13]. The first group (young group) consisted of 22 young adults in the age range of 18 to 43 years. Two experiments were performed on 10 of these subjects and only one on the remaining 12. The second group (elderly group) consisted of 15 elderly subjects in the age range of 64 to 84 years. No subject had evidence of metabolic, orthopaedic or neurological disorders, and the elderly subjects were all ambulatory and self-sufficient. Thirty-two experiments were performed on the first group and 15 on the second. Isometric contractions were sustained for 20 s at **1)** 20%MVC and 80%MVC and **2)** at supramaximal and submaximal stimulation currents, yielding respectively a maximum M-wave and a M-wave about 30% of the maximum. The two levels are respectively referred to as High Level Stimulation (HLS) and Low Level Stimulation (LLS). Stimulation frequencies of 20 Hz and 40 Hz were used. Stimulation was applied as described in greater detail in previous papers [7,8].

CV, MDF and MNF were computed on 20 one-second epochs for each contraction. The electrically elicited responses were averaged over each epoch. CV, MDF and MNF estimates were computed from the averaged responses. The time series of 20 values of each variable were fitted with a linear ($y = n - mt$) or an exponential ($y = ae^{-t/\tau} + c$) regression curve. The intercept of the curve with the y axis was taken as the initial value.

RESULTS

Voluntary Contractions: each of the 32 experiments of the young group showed an increase of CV, MDF, and MNF when the contraction level increased from 20%MVC to 80%MVC. This finding indicates that voluntary recruitment proceeds in order of increasing CV, resulting in increasing MNF and MDF, according to the size principle [1]. In the elderly group 12 out of

15 subjects showed an increase of CV and MNF values while only 8 showed an increase of MDF. Two subjects of this group showed a decrease of CV, MNF, and MDF when contraction level increased from 20% to 80%MVC. These findings suggest that recruitment does not always proceed in order of increasing CV in elderly subjects.

Stimulated Contractions: electrically elicited contractions torque was a small portion of MVC (below 30%), indicating that only a small portion of the muscle was activated by stimulation of a motor point. Four types of behavior were observed when stimulation current was increased from LLS to HLS. They are described in Table 1.

Table 1 Types of behavior observed when stimulation level is increased from LLS to HLS

Behavior types for stimulation increasing from LLS to HLS.	Number of experiments		
	Young group	Elderly group	Total
Type 1 (increasing MDF-MNF, increasing CV)	15	5	20
Type 2 (decreasing MDF-MNF, increasing CV)	8	4	12
Type 3 (decreasing MDF-MNF, decreasing CV)	6	1	7
Type 4 (increasing MDF-MNF, decreasing CV)	3	2	5
Total	32	12	44

When an experiment was repeated on the same subject it was not always classified in the same type, indicating the critical role played by electrode positioning. Fig. 1A shows the average initial values of CV in the two groups and in the 6 experimental conditions. Significant differences between the means of the CV initial values of the two groups (unpaired t-test) are not evident in any of the six experimental conditions. Differences become evident if paired tests are performed or individual increments. Fig. 1B shows the percent increment of CV when voluntary contraction level is increased from 20%MVC to 80%MVC, and when the electrical stimulation level is increased from LLS to HLS.

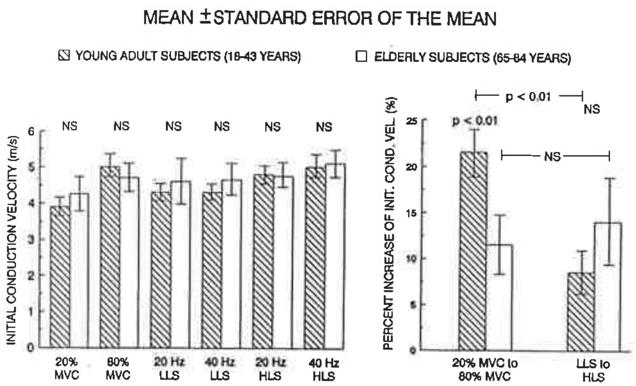

Fig.1 A) Mean value (± standard error or the mean) of initial CV in the 6 experimental conditions in the young group (white bars) and in the elderly group (dashed bars). NS means p ≥ 0.05 (t-test.
B) Mean value (± standard error or the mean) of the percent increase of CV due to voluntary contraction level increase from 20% to 80%MVC or to stimulation level increase from LLS to HLS.

DISCUSSION

Voluntary Contractions: increasing voluntary contraction level from 20%MVC to 80%MVC caused an increase of CV in 32 out of 32 exp. in the young group and in 12 out of 15 exp. in the elderly group. The elderly group shows an increase significantly smaller than the young group. In both groups the increase is significantly greater than zero.

This finding is consistent with histological observations indicating a smaller range of fiber diameters and a loss of large type II fibers in old age [6,10,11,12]. It is concluded that in voluntary contractions motor units are recruited in order of ascending CV, and that, in old age, either this order is altered or the range of CV values is smaller, or both. Spectral variables reflect this phenomenon in a less evident manner since they are affected by other factors, such as the tissue filtering function and the length of the surface potential distribution.

Stimulated Contractions: in 32 out of 44 exp. recruitment is still in order of ascending CV (type 1 and 2) while in 12 out of 44 exp. recruitment is in order of descending CV (type 3 and 4) suggesting an interaction between electrical excitability threshold of the motoneuron branches and their location in the current field. Changes of CV and spectral variables in opposite directions, consequent to an increase of stimulation intensity (see Table 1), may be due to the combined effect of the motor unit recruitment order and of the different tissue filtering function associated with the geometric location of the recruited motor units within the muscle. The recruitment of more superficial motor units would lead to an increase of MDF and MNF because of the higher cut-off frequency of the tissue filter. The change of average CV may enhance, reduce of reverse such increase according to the CV value of the newly recruited motor units. The recruitment of progressively deeper motor units would lead to a decrease of spectral variables, even if the average CV would increase. These results suggest that, in general, more excitable nerve fibers, innervating motor units with higher CV, may be located more deeply and, therefore, require higher current levels for excitation. This finding is in agreement with histological observations [9].

CONCLUSION

It is concluded that in voluntary contractions motor units are recruited in ascending order of CV, and that, in old age, this order is occasionally reversed while the increase of CV is generally smaller reflecting a more uniform motor unit population. It is also concluded that during surface electrical stimulation of a motor point, motor units of the tibialis anterior are recruited in ascending order of CV in 72-75% of the experiments. This percentage is not affected by age. Also the increase of CV when stimulation level is increased from LLS to HLS is not affected by age. Since the muscle properties mostly affected by age are the number and size of motor units, the percentage of type II fibers, and the size of the fibers, it may be concluded that such changes may be more properly detected by comparing myoelectric signal variables obtained from low and high level voluntary contractions rather than by electrically elicited contractions.

Spectral variables are affected by CV changes as well as by the location of the active motor units, and do not necessarily change in the same direction of CV when stimulation level is increased. Electrode positioning appears to be a critical factor, affecting repeatability in the same subject. It appears that the greater degree of mechanical fatigue observed during Electrical Stimulation with respect to that observed during voluntary contractions cannot be attributed to a reverse recruitment order of the muscle motor units. Finally, the increase of CV consequent to increase of voluntary contraction level from 20% to 80%MVC appears to be a good indicator of muscle fiber type constituency. This finding, associated to the rate of decay of CV and spectral variables may provide a basis for non-invasive muscle characterization.

Acknowledgment

We express our gratitude to L. LoConte, M.V. Actis and U. Massazza for their help in this work. This work was performed within the framework of a scientific cooperation between the NeuroMuscular Research Center of Boston University and the Politecnico di Torino, Italy, involving the U.S.S.L. 46 of Santhia'(VC), Italy. Partial support was provided by Liberty Mutual Insurance Company, by the Italian Ministry for University and Scientific Research and by the National Research Council of Italy.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

WHICH PARAMETERS AT MOTOR UNIT LEVEL INFLUENCE THE MEDIAN FREQUENCY OF THE SURFACE EMG.

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INTRODUCTION

Presently one of the parameters most often used in surface EMG investigations is the median frequency, MF. It is used to assess changes on motor unit level that occur in neuromuscular disorders[3,5] and it is often used in studies dealing with muscle fatigue[2]. Recently it has been shown that it may also be used to monitor recruitment strategies[7].

Although the MF is often used, its relation with parameters at motor unit level is still not quite known. These parameters can be subdivided in MUAP characteristics (e.g. shape, amplitude and duration of the MUAPs), parameters describing the individual firing process and synchronization parameters. We investigated the sensitivity of the MF for these parameters in a simulation study and determined their possible contribution to observed MF changes by relating this information to their physiological range.

METHODS

The simulation model assumes that the surface EMG signal can be described as a linear summation of a large number of action potential trains. Each train represents the electrical activity of one motor unit.

Furthermore it assumes that a motor unit action potential train (MUAP train) can be generated by convolution of a train of delta pulses with a MUAP shape [,6]. The train of pulses simulates the firing of the motor unit. The MUAP represents the motor unit action potential as it is recorded with the surface electrodes.

Instead of using a single synthetic shape of the MUAP a library has been created in which a number of MUAP shapes of standardized amplitude and duration are present. These may be derived from experimentally obtained or from artificially created MUAPs. In this way a large flexibility is created to study various effects of the MUAP shape on the EMG patterns.

In order to obtain a MUAP for a specific train a library shape is chosen and multiplied with both a time and an amplitude factor. These factors are drawn at random from two Gaussian distribution functions in order to simulate the wide range of amplitudes and durations of MUAPs. So within a train the MUAP's shape, amplitude and duration remain unchanged, but these parameters differ for different MUAP trains.

For each train the mean firing frequency is drawn at random from a Gaussian distribution function to simulate the different mean firing frequencies of the individual motor units. Within each train the subsequent values for the inter pulse interval time IPI are drawn from another Gaussian distribution function to simulate irregularities in the firing process.

Synchronization is implemented by placing around each firing moment of one generated MUAP train a Gaussian distribution function[8]. From this function the firing moments of the synchronized trains are drawn. Both the width of this distribution function SV and the number of trains that show synchronized activity NS can be chosen freely.

RESULTS

The most relevant results from the sensitivity analysis are listed below.

Influence of the individual MU firing processes

Mean firing frequency

The mean firing frequency has only a small influence on the value of the MF.

The MF increases about 5% when the mean firing frequency changes from 5 to 20 Hz, the normal range.

Regularity of firing

The MF increases only slightly when motor unit firing becomes less regular. MF increases less than 5% when the standard deviation of the IPI distribution function varies from 5% to 20% (of the mean IPI).

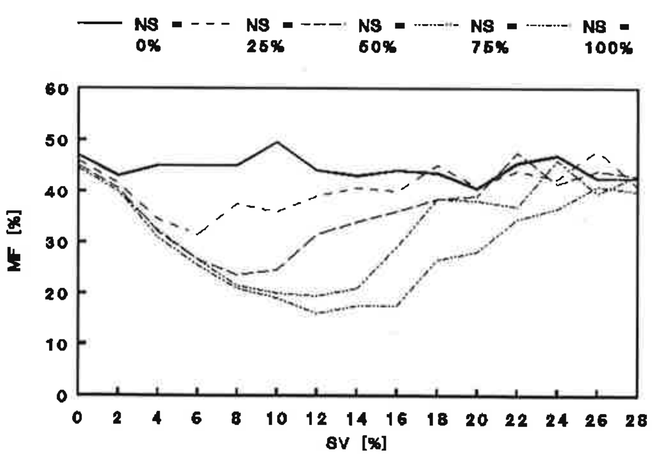

Fig. 1: The influence of the level of synch onization on the MF.

Influence of synchronization

Fig. 1 shows the influence of synchronization on the MF for different parts of the population of motor units firing synchronized. In this figure MF is plotted as a function of SV for different values of NS. It is clear that changes in SV affect the MF to an extend depending on the value of NS.

Influence of the MUAP characteristics

MUAP duration

Fig. 2 shows the relation between the reciprocal value of the MUAP risetime RT and the MF for 3 different MUAP shapes. It can be seen that the MF is linearly related to the reciprocal value of the mean RT.

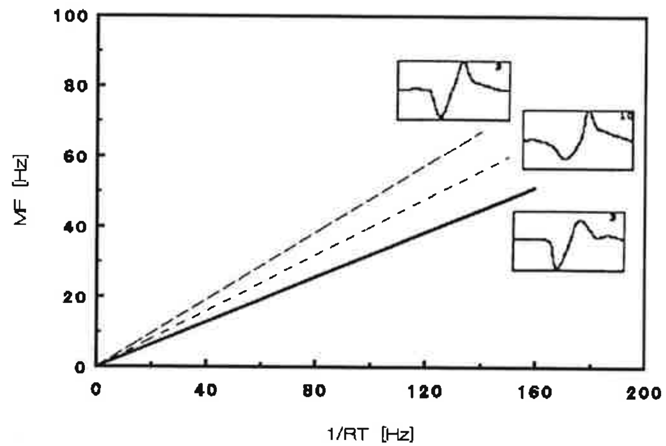

Fig. 2: The elation between MF and RT and the influence of the MUAP shape.

Variability in MUAP-duration

It was found that the amount of variation in RT has only a very small influence on the MF. MF increases less than 3% when the standard deviation of the RT distribution function varies from 10% to 25% (of the mean RT).

MUAP shape.

Differences in MUAP shapes are reflected in the relation between MF and the reciprocal value of RT (Fig. 2). This relation was found to be linear with zero intercept for all shapes. However the slope of the relation S differs for different MUAP shapes. (This can be interpreted as a sensitivity change in MF for RT).

A realistic estimate of the practical importance of this effect was made using MUAPs extracted from surface EMG recorded under normal voluntary contraction by means of a needle triggered averaging method[4]. It was found that MUAP shape differences can play a role in intra individually observed MF changes. It was also found in inter individual comparison that the average S value did not differ significantly.

DISCUSSION

From the sensitivity analysis combined with information on the physiological range of the examined parameters evolved that three parameters potentially have a dominant influence on the MF of the surface EMG signal. These are: the shape and the mean RT of the MUAP and the level of synchronization.

This implies that changes in the MF, e.g. observed at different contraction levels or during fatigue, can only be caused by changes in muscle fiber conduction velocity, in recruitment level or in level of synchronization. It also implies that the individual firing behavior of unsynchronized motor units has only a neglectable influence on the MF.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

FINE MOTOR CONTROL IN NORMAL MUSCLES OF THE HAND AND REGULARITY IN THE SPEED OF VOLUNTARY WRITING

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INTRODUCTION

The initiation of a basic method to evaluate *the fine motor control* with the use of the EMG audio-visual biofeedback of single motor unit activity, served to various fundamental and clinical studies. Of great concern, factors, as *movements*, influencing the chosen single motor unit maintenance in pure activity, demonstrated the existence of different levels of motor ability in healthy subjects (1). A step learning procedure of maintaining a single motor unit even during *utilitarian* motions, has thereafter been developed (2). However, a practical and complementary approach in this area is necessary. The electromyography of the influence of fine motor control of muscles of the thumb on the *speed* and *duration* and on regularity of *writing* motion should give beneficial physiologic results.

MATERIAL AND METHODS

Fourteen healthy *subjects*, 6 males, 8 females, aging from 20 to 40 years old participate. Three muscles of the right thumb, differently innervated, are studied. Within these, the *Abductor Pollicis Longus* (APL) is selected for the present analysis. The APL, summit territory, is reached with a bipolar fine wire electrode, half way between the lateral epicondyle of humerus and the prominence of head of ulna. The territory understudy is constituted of small muscle fibers of different oblicities which should act in delicate coordinated movements of the thumb (3). An adapted *multiorthotic* device has been developed to insure the *body stability* and to prevent undesired movements. Subjects are installed in a half-lying position, their feet rest on an adjustable platform. The right upper limb is placed on a molded plexiglass adjustable device. This, allows 15 degrees abduction and 45 degrees flexion of the arm. The forearm and hand are placed in natural pronation and extension. The third to fifth fingers are immobilized in a relaxed state of flexion. The thumb and index hold a pencil on a shelve in preparation for writing a line downward. This is called the "O" posture. A special *electrogoniometric* (EGM) system developed includes a potentiometer. The probe of the system is adjusted at the dorsal distal segment of the first metacarpal bone in line with flexion-extension of the carpo-metacarpal (C-M) joint. A cursor

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adapted to a ruler, identify the "O" posture of the thumb and any degree of C-M displacement. The EGM recording is made simultaneous to the EMG traces on a tape recording system. Three low level *EMG* pre-amplifiers are utilized at a bandwith from 5 Hz to 10 KHz and at a gain of 10 K. After testing the electrodes placement, ten *tests* are given within approximately one hour. With the aid of the audio-visual biofeedback signs of the EMG of APL, subjects learn to isolate and maintain in repetitive activity a single motor unit during two minutes at the "O" posture (test 1). Subjects are told to hold the finest motor control in APL during a "slow" writing a line downward (applied concentric contractions) of five degrees C-M extension-abduction (test 2). The slow speed value (1-4s) is left to the subject will. At the downward point (test 3), subjects are asked to hold the fine motor control or to recall it. During the writing upward (test 4) and at the "O" posture (test 5), subjects are asked to continue to hold or recall the single motor unit previously chosen. The fine motor control is also required before (test 6), during "fast" (1s) writing a line downward (test 7) and upward (test 8). Subjects are asked to hold the chosen single motor unit at the "O" posture (test 9). The test 10 is given during the single motor unit responses evoked rhythmically, by group of contractions, during fast pressures of the thumb on pencil. This facilitate active stretching of the understudied APL. The APL myoelectric observations are read on a storage oscilloscope at a high speed (5 ms/D) to qualify the *levels of fine motor responses* and recorded on linographic paper at 2"/s to calculate the duration and speed of motions. Perfect isolation of the chosen motor unit is called "Very Good". Predominance of the single motor unit action, transfer to another single motor unit action and inhibition of activity are qualified as being a "Good" control. Light activity is called a "Fair" fine motor control. For each test, the observed quality of levels of control is tabulated. The frequency distribution of the different success, the $\overline{X}$ and the regularity (Sd) of the duration and of the calculated speed are analysed.

RESULTS

The APL can produce different *degrees of fine motor control* in each test applied (Fig. 1). A Very Good fine motor control is obtained most frequently at the initial upwright posture (71%). Several (29%) skillful subjects established this Very Good success even during the writing a line downward which require a concentric type of contraction in APL. The decrement of success is mostly marqued in fast motion, where 9% of subjects held their chosen motor unit in activity. It is also during fast writing downward that a Fair success is more frequently obtained (64%). Very Good motor ability is even sustained in 31% of subjects during the active rhythmic test, stretching the APL. Slow and fast writing a line downward and upward affect the quality of fine motor control achieves at the "O" posture.

In a few subjects (3/14), even if the single motor unit control is lost during the rapid writing activity, this control is seen to be quickly recall just after motion. The increment of effort to maintain fine motor control during both type of contractions, slow down the $\overline{X}$ of the *speed* of motion significatively while doing the last 4th and 5th degrees of movements ($\overline{X}$= from 0,76 to 0,30 m/s in concentric and, from 0,97 to 0,57 m/s in eccentric). However, the regularity of the speed of the concentric type of contraction is improved at these last degrees of motion (Sd= from 1,25 to 0,28 m/s). The $\overline{X}$ *duration* of movement slows down ($\overline{X}$= from 2,12 to 3,93 s) in increasing the degrees of motion of the applied concentric contraction. The regularity of the duration of applied concentric contraction is particularly affected at the last two degrees of motion. In the applied eccentric contractions, the total duration of slow movements is more regular (Sd= 5,17 s) than in concentric contractions (Sd= 7,68 s). In analysing the individual variable values, again a higher regularity within the speed (Sd= 0,88 m/s versus 1,14 m/s in concentric) and duration (Sd= 0,95 s versus 2,10 s in concentric) of eccentric contractions is well observed. The willed effort to maintain the finest motor control in active repetitive rhythmic stretching of APL, facilitates intermittent muscular responses of different levels of motor control (Fig. 1, test 10). Single motor unit is often activated in pair of potentials and *doublet* discharges are frequently seen (50%). In one subject, the shape of the first of certain pairs of discharges shows an *additional myoelectric deflection* at the terminal phase of slow and fast muscular eccentric contraction (Fig. 2). From the foregoing observations of this special similar shape deflecting signal, it became evident that a functional affiliated intermittent myoelectric responses can be made active at always an exact same timing of the terminal phase of muscular contractions and is always linked to the first potential of the pair of discharges (Fig. 2).

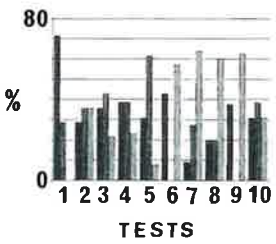

Fig. 1

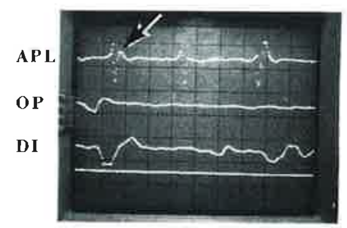

Fig. 2A

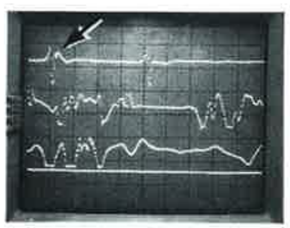

Fig. 2B

Fig. 1. Levels of Fine Motor Control: ■ Very Good; ■ Good; Fair.

Fig. 2. Possible Alpha-Gamma Linkage in APL. Arrow: Possible muscle spindle EMG.
A-During active slow upward motion; B-During active fast upward motion (S14).
Cal. A: 200μV= 2 Vertical lines; D: 5 ms/D.

DISCUSSION AND CONCLUDING REMARKS

The different levels of fine motor control that can be achieved in a subject APL muscle at a same "O" *posture* illustrate the existence of disparate motor programs for achieving a particular performance. This is in accord with a recent review of motor programming theories (4). It is normal that the level of concentration and the willed effort magnitude vary during an experimental session, therefore, these can distort the quality of motor control. The different levels of fine motor control utilized by young adults to perform *various speed*, as regular as possible, express the necessary value of these tests. We know, today, that slow and fast movements do require different afferent nervous pathways to be effective (5). The present measures indicate further, that fine motor control can be obtained in APL but is significatively affected by fast writing downward concentric contractions. We believe that rapid velocity involves an increment of reflex action which in turn may restrains the purity of higher system activity. This do not go against the theory that fast motor behavior is largely programmed in the central nervous system (5). Finally, the use of active *rhythmic* muscular stretching (test 10) brought in light the predominance of a well organized central nervous system which act possibly on the alpha and gamma motoneurones simultaneously, and then, on the foregoing same motor unit under its own myotatic reflex action. The significance of this original observation deepens the theoretical value of the spindle timing reflex action in voluntary movement. This complementary approach and the actual results from APL skilled performance, open a hope to improve the evaluation of the fine motor control utilitarian ability in EMG.

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- Research supported by FRSQ and CNPq of Brazil. - Photographies are from Mr. J. Léveillé, Département d'Anatomie, Université de Montréal.

Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

EXAMINATION OF SEGMENTAL REFLEX PATHWAYS IN INTACT HUMANS

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INTRODUCTION

The tendon-tap reflex has been used to identify both training (1-3) and aging (4-5) adaptations of the human neuromuscular system, as well as in the diagnosis of various neuromuscular disorders (6). Whereas it is clear that the muscle response to stretch is not confined to the homonymous musculature, perhaps a better and more comprehensive understanding of segmental influences acting on the reflex arc can be gained by examining conditioned reflexes.

The purpose of this study was to examine human quadriceps and triceps surae excitability following a conditioning stimulus to either the ipsilateral or contralateral limb. By examining the reflex recovery profiles of the patellar and Achilles tendon-tap reflexes, the exact time-course of motoneuron excitability changes caused by a mechanical conditioning stimulus could be determined. Also, it was the purpose of this study to examine the spinal connections mediating the human quadriceps and triceps surae muscles.

METHODS

Data were obtained from 12 subjects (mean age = 28.4 yrs) who read and signed a subject consent form. These individuals were drawn from a healthy population with no known history of neurological, orthopedic or neuromuscular disorders. Using electromagnetic solenoids capable of monitoring the tendon tap force, four experimental conditions were randomly assigned on separate test days: 1) the patellar tendon-tap reflex (PTR) was conditioned by a tap to the contralateral patellar tendon; 2) the Achilles tendon-tap reflex (ATR) was conditioned by a tap to the contralateral Achilles tendon; 3) the ATR was conditioned by a tap to the contralateral patellar tendon; and 4) the ATR was conditioned by a tap to the ipsilateral patellar tendon. The conditioning stimulus preceded the test reflex by 25, 40, 55, 70, 85, 100, 115, 130 or 145 ms. Control responses, during which tendon reflexes were elicited in the right leg without a preceding conditioning stimulus, were also obtained. Three trials were randomly administered at each conditioning interval, for a total of 30 reflex trials per experimental condition. Changes in test reflex excitability were determined by examining both electromyographic and force-time characteristics of the reflex response. Bipolar recording electrodes of 1 cm diameter were positioned over the belly of the gastrocnemius when examining the ATR, and over the belly of the rectus femoris when examining the PTR. A two centimeter intraelectrode distance was used, with the ground electrode positioned midway between the two recording electrodes. Data was collected online with a microcomputer equipped with a data acquisition board, with the sampling rate set at 2 kHz. On each trial, the following dependent measures were examined: peak isometric force, EMG latency, electromechanical delay, force latency, integrated EMG, peak to peak EMG activity, contraction time and half-relaxation time.

RESULTS

PTR: When the PTR was conditioned by a tap to the contralateral limb, the force output of the reflex was significantly enhanced. Maximal facilitation occurred at the 145 ms interval, when peak force was 152.3 percent above the control value. Similar results were found for the EMG data, and these recovery profiles are shown in Figure 1. The analysis of the trend components for peak force demonstrated a significant linear trend. Facilatory changes were also noted for the following dependent measures: force latency, electromechanical delay and EMG latency, with a significant lengthening of contraction time.

ATR: When the ATR was conditioned with a tap to the contralateral Achilles tendon, the reflex force was slightly facilitated at the 40, 55 and 70 ms conditioning intervals. This was followed by a long-latency inhibition in reflex force, as shown in Figure 1. Maximal facilitation was at the 40 ms conditioning interval (119.2%) whereas maximal inhibition occurred at the 130 ms conditioning interval (93.4%). The reflex recovery profile for peak force was characterized by a significant cubic trend. The EMG data displayed a similar profile, as shown in Figure 1. No changes were noted for contraction time, force latency, EMG latency or half-relaxation time.

When the conditioning stimulus was a tap to the contralateral patellar tendon, a quartic trend in the peak force recovery profile was uncovered, as reflex force was returning to control values at the 145 ms interval. This can be seen in Figure 1. The conditioning tap again caused a slight and early facilitation of the ATR that was followed by a general, but slight inhibition. Significant increase in the force output of the reflex was noted at the 55 ms (116.0%) and the 70 ms (120.7%) conditioning intervals. As in the other experimental conditions, integrated EMG displayed a similar profile (Fig. 1). Whereas no changes were noted for EMG latency, contraction time or half-relaxation time, force latency and electromechanical delay displayed the early facilatory effect.

When the conditioning stimulus to the ATR was a tap to the ipsilateral patellar tendon, a cubic trend for peak force was again uncovered, with a mild and brief facilatory phase followed by a more distinct inhibitory phase. In this condition, maximal facilitation occurred at the 40 ms interval (120.7%) whereas

maximal inhibition occurred at the 100 ms interval (65.8%). No changes were noted for force latency, contraction time and half-relaxation time, while electromechanical delay exhibited a short-latency facilitation and EMG latency displayed a long-latency inhibition.

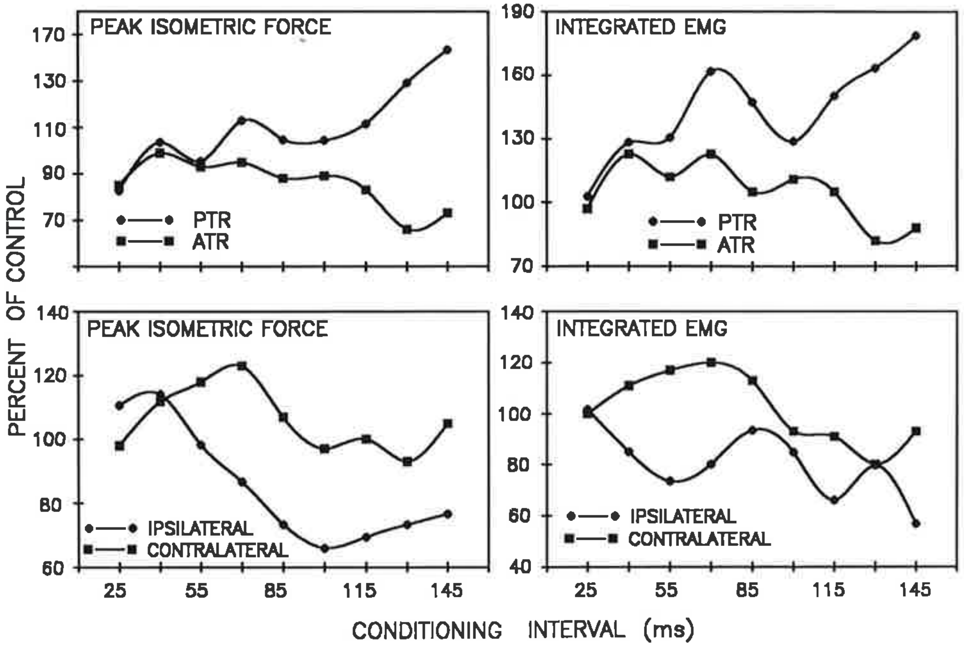

Fig. 1. Reflex recovery profiles for peak isometric force and integrated EMG. PTR: patellar tendon-tap reflex conditioned by a tap to the contralateral patellar tendon; ATR: Achilles tendon-tap reflex conditioned by a tap to the contralateral Achilles tendon; Ipsilateral, Contralateral: Achilles tendon-tap reflex conditioned by a tap to the ipsilateral and contralateral patellar tendon, respectively.

DISCUSSION

These results demonstrate that a mechanical conditioning stimulus produces changes in reflex excitability of both the quadriceps and triceps surae muscles in intact humans. These results also demonstrate that a contralateral conditioning stimulus produces a different recovery profile for the PTR when contrasted with the ATR.

In regards to the motoneuron excitability changes noted, the role of cutaneous receptors must be considered. In young and old adults, ipsilateral and contralateral cutaneous input to the quadriceps and triceps surae muscle groups have been shown to have facilatory effects on the force output of the patellar tendon-tap reflex (7). Similarly the quadriceps muscle has been shown to be facilitated by cutaneous skin stimulation to various skin areas (8).

Clearly, other segmental mechanisms may be operating as well. For instance, muscle spindle discharge characteristics (9) and polysynaptic spinal connections (10) might be considered as potential mechanisms mediating these changes. Similarly, the role of supraspinal pathways cannot be overlooked (11).

An interesting observation from these data is the quadriceps muscle displays a different recovery profile than the triceps surae muscle when conditioned by a contralateral tendon-tap. While differential effects have been reported using ipsilateral homonymous conditioning (12), these data support the notion that a crossed-spinal mechanical conditioning stimulus causes different motoneuron excitability changes for the quadriceps and the triceps surae.

These results also demonstrate that the conditioned ATR profile is distinctly different from that characteristically obtained with the electrically evoked H-reflex (13). These differences may be explained by the differences in the spatial and temporal characteristics of the conditioning stimulus, as well as the involvement of cutaneous receptors (14).

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

CUTANEOUS AND MUSCLE AFFERENT INPUTS TO HUMAN TRICEPS SURAE ALPHA MOTONEURONS

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INTRODUCTION

Afferent signals from muscle, skin and joints interact with descending motor commands to modulate the output of the alpha motoneuron pool. Details regarding these interactions are lacking, although recent hypotheses concerning the sharing of common interneurons[1] provide a new, integrative approach to the study of neuronal pathways. Investigation of the interaction of reflex and descending pathways necessitates an understanding of the potential effects individual inputs may have on the alpha motoneuron pools. In man, most information on the central connections of peripheral afferents has been gathered for muscle afferent inputs onto lower limb spinal motoneurons. Cutaneous pathways have been evaluated primarily from the perspective of responses to noxious stimuli. Experiments here have focused on describing the sign, latency, and potency of effects of non-noxious cutaneous and muscle afferent stimulation on the discharging of motor units in the three heads of triceps surae.

MATERIAL AND METHODS

A peristimulus time histogram (PSTH) technique[2] was used to cross correlate the discharging of a single unit motor unit with the delivery of electrical stimuli to the sural or peroneal nerves. A two step procedure for validating and establishing significance of an effect consisted first of imposing parabolic limits (2.58 standard deviations) onto an integrated transformation (cumulative sum plot) of the PSTH[3]. If a potential effect touched or broke through the limits, a Z-score test of significance ($p<.01$) was performed[4]. For peroneal stimulation, 244 spike trains were obtained from 10 subjects. For sural stimulation, 168 spike trains were obtained from 14 subjects. Stimulation consisted of either single, 0.1 ms pulses, delivered at 3/s, or in some instances with sural stimulation, 5-15 ms, 300/s trains, at 3/s.

RESULTS

Sural Nerve Stimulation Effects

Reflex effects onto motoneurons from the three heads of triceps surae displayed similar as well as differential effects (Figures 1 & 3). The similarities included short latency inhibitions (I_1 effects with onset latency between 28 and 55 ms), long latency inhibitions (I_3: 90-145ms) and a broad intermediate range of excitatory effects (E_2: 40-135 ms). Infrequent, short latency (E_1: 28-34 ms) excitations most often occurred in Soleus (6.4% of all Soleus responses). LG units displayed an intermediate inhibition (I_2: 60-90 ms) that was seen only twice each in MG and Soleus.

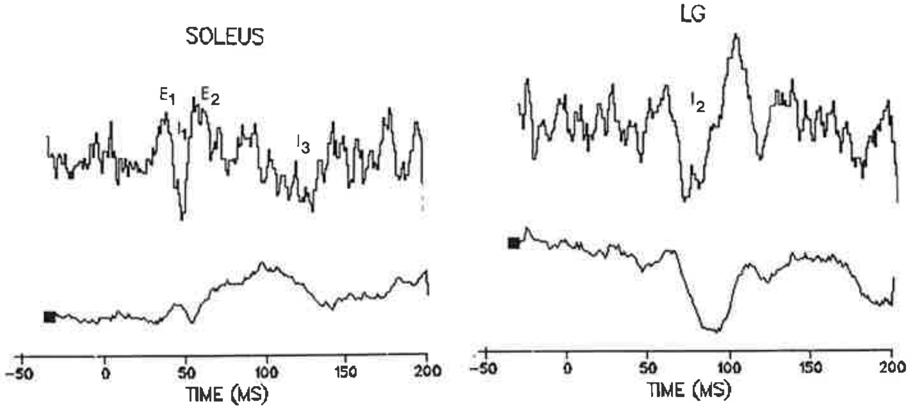

Fig. 1. Characteristic short (E_1 & I_1) and long latency (E_2 & I_3) reflex effects for sural stimulation as exemplified in a Soleus motor unit. An additional intermediate inhibition (I_2) was also observed primarily in LG motor units.

Peroneal Nerve Stimulation Effects

Peroneal nerve stimulation effects were assessed only for Soleus motor units. A short latency inhibition (28-50 ms) and secondary excitation-inhibition complexes (50-100ms), similar to those seen with sural stimulation, predominated in most units recorded (Figure 2a). In addition, an exceptionally potent early excitation (28-48 ms) was seen in 18% of the units (Figure 2b) which always preceded the more consistent short latency inhibition.

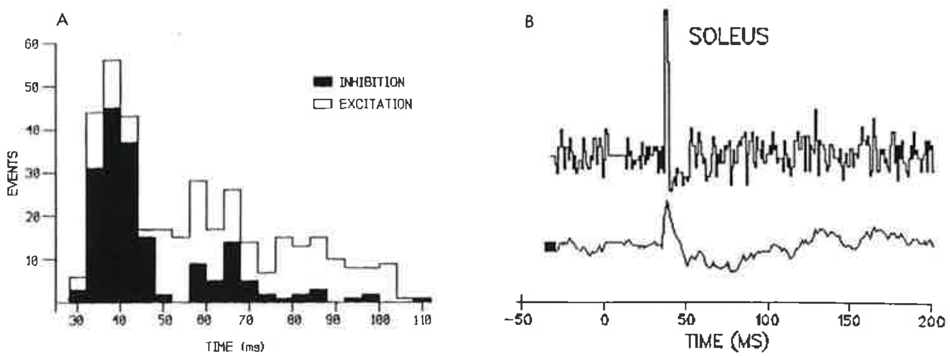

Fig 2. A: Summary of onset latencies for peroneal stimulation effects indicates a predominance of short latency inhibitions. B: Representative example of short latency excitatory effect with peroneal stimulation.

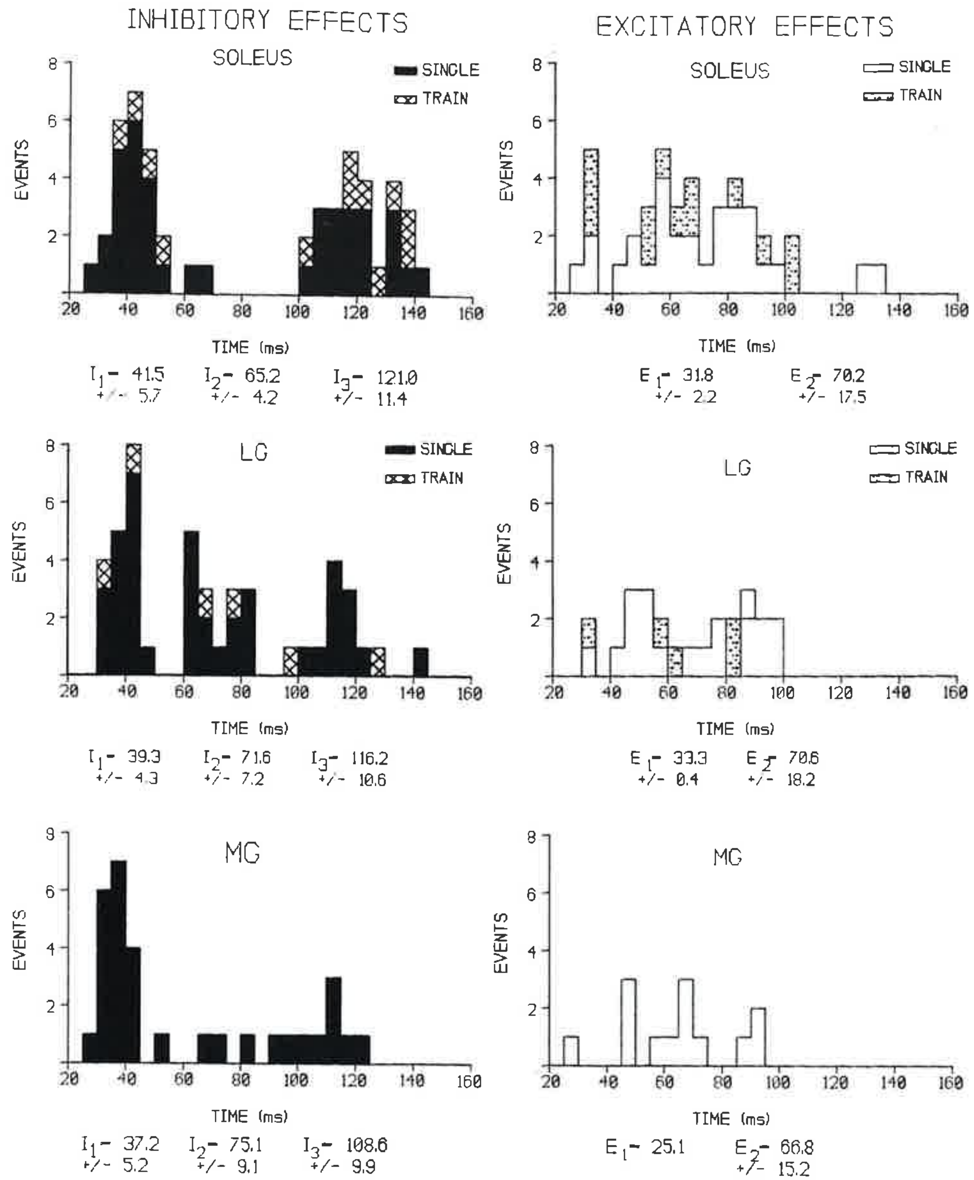

Fig 3. Summary of onset latencies for sural stimulation effects in three heads of triceps surae. For Soleus, 50 of 69 (75%) units displayed effects; in LG, 45 of 55 (82%) showed effects and for MG, 27 of 33 (82%) showed effects. The percentage of excitatory:inhibitory effects were 42:58 for Soleus, 28:72 for LG and 29:71 for MG.

DISCUSSION

The results from the sural stimulation experiments reveal that the response of triceps surae motoneurons is more complex than singular excitatory or inhibitory effects. The predominance of an effect is likely determined by the specific motor task required. That is, the full expression of either an excitatory or inhibitory response is likely due to the interaction of the descending motor command with peripheral afferent signals. An example might best illustrate this point.* Stimulation of the posterior tibial nerve at the ankle evokes a surface response in Soleus that closely resembles the complex reflex effects reported here for single units. Under static conditions, the evoked response is primarily a long duration inhibition (latency about 50 ms) followed by a secondary excitation. When evoked during cycling and while a steady background level of Soleus activity is maintained (10% MVC), the reflex response is modulated according to the demands of the task. When the limb is moving to develop an acceleration to the bicycle crank, excitatory effects are introduced into the period of inhibition. This reflex reversal is appropriate to assist in delivery of the needed Soleus torque. The reflex responses described here undoubtedly reflect the rudimentary pathways over which these excitatory and inhibitory effects are conveyed.

The functional implications of the two major differences in reflex effects (greater percentage of inhibitory effects in LG and MG compared to Soleus, and the predominance of an intermediate inhibition in LG) are not readily apparent. The shift from Soleus to LG activation seen for human lengthening contractions[5] may be due to a depression of these inhibitory LG effects by a change in the descending command when the task shifts from that of a shortening to a lengthening contraction. Such a possibility awaits experimental confirmation.

Finally, the common reflex effects to both sural and peroneal stimulation suggest a wide range of future experiments which may begin to explore the possible sharing of interneurons common to these segmental pathways.

ACKNOWLEDGMENTS
This work was supported by PHS Grant NS 24991.

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*Adapted from the PhD thesis of David A Brown, The University of Iowa, 1989.

Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

EMG POWER SPECTRUM ANALYSIS OF PARASPINAL MUSCLES: THE EFFECT OF BODY MASS AND SEX

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INTRODUCTION

EMG power spectrum analysis investigations of paraspinal muscle function have shown that the recording procedure is reliable (1), and can be used to differentiate between normal controls and subgroups of chronic back pain patients (2,3), with patients showing higher initial median frequencies and greater shifts to lower frequencies (muscle fatigue)(2). The findings were interpreted to reflect a deficient endurance capacity in the patient group indicative of variations in muscle fiber properties of paraspinal muscles.

Although a close relationship between the myographic parameters of the power spectrum and the histological composition of muscles has been proposed, evaluations of spectral changes in paraspinal muscles with regard to variables related to muscle histology such as physiological anthropometric measures are still lacking. This is a potentially important issue in EMG power spectrum research with pain patients since findings of differences in muscle strength or endurance may be attributable to variations in anthropometric characteristics (4,5).

Unfortunately, the association between power spectrum parameters and anthropometric measures has not been investigated beyond the use of body weight to normalize load level (i.e. expressing load as the mean trial force to body weight ratio, instead of "maximum voluntary contraction")(6).

Similarly, sex differences in EMG power spectrum parameters of the paraspinal muscles have not been studied. Although there is some information available regarding histological differences in paraspinal muscle tissue between men and women (7), with women showing smaller Type II muscle fibers in the superficial and deeper paraspinal muscles than men, investigations to date have either used male subjects only (3), or have not examined sex effects in their data (2).

The present study was undertaken to investigate if EMG power spectrum parameters of paraspinal muscles are sensitive to effects of body mass and sex.

METHOD

The data of 63 subjects who participated as control subjects in an ongoing back pain investigation (2) were analyzed. Details of the constant force contraction and EMG recording procedure employed have been published elsewhere (1). Subject characteristics including the Body Mass Index ($BMI=weight/height^2$) are summarized in Table 1.

TABLE 1

	Male	Female
N	30	33
Age (yr)	27.5 (8.6)	29.6 (9.2)
Height (m)	1.81 (.05)	1.65 (.06)
Weight (kg)	79.04 (8.1)	61.6 (8.9)
BMI	24.0 (1.6)	22.7 (2.7)

Estimates of the initial median frequency (MF) and muscle fatigue rates for the iliocostalis and multifidus muscles were obtained through linear regression analyses. Subsequently, multiple regression statistics were used to evaluate the effect of body mass and sex on the EMG power spectrum parameters of the iliocostalis and multifidus muscles.

RESULTS

Table 2 summarizes the multiple regression findings. The results indicate a differential and moderate influence of BODY MASS and SEX on the EMG parameters of the two muscle sites.

TABLE 2

	t	p	sR^2
Multifidus			
MF ($F=7.20; p<0.002; R^2=0.199$)			
BMI	-3.71	0.001*	0.19
Sex	2.01	0.049*	0.01
Fatigue rate ($F=0.81; p<0.450; R^2=0.027$)			
BMI	0.98	0.329	0.02
Sex	0.43	0.667	0.01
Iliocostalis			
MF ($F=3.65; p<0.032; R^2=0.110$)			
BMI	-2.70	0.009*	0.11
Sex	1.08	0.285	0.00
Fatigue rate ($F=7.82; p<0.001; R^2=0.213$)			
BMI	1.30	0.197	0.02
Sex	3.09	0.003*	0.19

* significant ($p<.05$)
sR^2 : unique variance

The index of BODY MASS accounted for almost all variation in MF values of the multifidus and iliocostalis, with the Pearson correlation coefficient showing an inverse relationship between the two variables at both muscle sites (multifidus $r = -.38$; iliocostalis $r = -.28$). However, no statistically significant relationships were found with regard to fatigue parameters of the iliocostalis and multifidus.

SEX accounted for almost all variation in the regression equation of the fatigue rate of the iliocostalis, with women displaying elevated muscle fatigue (females = -9.87; males = -5.06). Although SEX was also a significant factor in estimating the MF of the multifidus (females = 119.9 Hz; males = 122.4 Hz), it only accounted for minimal unique variance.

DISCUSSION

The present observation of body mass and sex-related differences in the EMG power spectrum profile suggests that the EMG recording procedure is sufficiently sensitive to respond to these factors.

Body mass showed a negative relationship with initial median frequency in the multifidus and iliocostalis muscles, which was independent of sex. This suggests that anthropometric factors should be considered when comparing groups within a single study, and between different studies. For example, using the index of body mass as above, Roy et al. (3) studied disparate patient and control groups (26.8 vs. 24.7), which may have confounded the outcome and interpretation of their investigation. Based on these group BMIs, their patients would be considered overweight according to Canadian Guidelines for Healthy Weights (8)[1]. Similarly, comparisons between studies would be complicated by the fact that Roy et al.'s groups differ substantially from the subjects used by Biedermann et al. (2) on the BMI (patient = 21.4; control = 21.1).

With regard to the effect of sex, women displayed higher fatigue rates than men in the iliocostalis muscle. These findings seem to be consistent with the observation by Bagnal and colleagues (7) that, on the basis of histological analysis, women had a lower strength factor than men in superficial back muscles. Since sex emerges as an identifiable factor in EMG power spectrum research of paraspinal muscles, it has important implications for future research projects because results from male subjects (which predominate the relevant literature) may not be generalizable to women. Similarly, when mixed samples are used, sex should be treated as an independent factor.

Thus, this investigation has indicated that sex and anthropometric factors may influence significantly EMG power spectrum parameters. The index of body mass used in this regard, however, may not represent the best measure of muscle mass and potential strength/endurance, since it is based on total body weight, including body fat. This may interact with the effect of sex since women usually have more body fat than men, incorporating more "dead weight" into their index of body mass. It is suggested that future studies include anthropometric measures more directly related to muscle mass, such as lean body weight (9) in order to assess its confounding effect more closely.

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[1] Canadian Guidelines for Healthy Weights (1988) recommended that the following zones for the Body Mass Index be used to evaluate weight for adults aged 20 to 65 years:
(a) <20 : may be associated with health problems for some people
(b) 20 - 25 : good weight for most people
(c) 25 - 27 : may lead to health problems in some people
(d) >27 : increasing risk of developing health problems

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

NON-LINEAR PHASIC DYNAMICS OF MUSCLE ACTIVATION AND FORCE GENERATION

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Abstract

A model has been developed of the dynamics of the excitation of the motoneuron, the transduction of action potentials to muscle activation and the generation of muscle force. The PEXA model has been calibrated and tested against in vivo experimental data and reproduces the non-linear adaptation behavior of the moto-neuron and the non-linear enhancement of muscle tension at short inter-stimulus intervals. The model will be useful in detailed dynamical modeling of high speed voluntary movements.

Introduction

When we make our fastest possible movements, as in emergency reactions or sporting events, muscles are driven by bursts of neural excitation as short as 50 ms. (as manifested in EMG bursts). For the quantitative analysis of these movements, it is essential to include dynamics which occur at these time scales. For practical reasons, most experimental data and models of neuro-muscular interaction are concentrated on slow or tonic contractions.

A new model has been developed of the dynamics of moto-neuron firing rate adaptation and the response of muscle activation and the resulting force development.[1] The "Phasic Excitation-Activation," (PEXA) model (Figure 1) focuses on the quick increase and then decline of firing rate seen in moto-neurons driven by steps of depolarizing current, and the dynamics of muscle activation when driven by neural impulse trains. The model parameters have been identified from experimental data in the literature. The model consists of three components which represent the moto-neuron, the mucle activation process, and the mechanics of force generation.

Excellent experimental data, primarily in cat muscle, describes the short time dynamics of these processes. These basic experiments measured responses of motoneurons when driven intracellularly by steps of transmembrane depolarizing current,[2,3] the force output of motor units when their axons were stimulated with different patterns of pulses,[4] and the concatenation of these two paradigms, the force response of motor units in vivo to a step or ramp of depolarizing current injected into the controlling motoneuron.[5,6] We thus can draw on experimental data which give the behavior of the individual components[3,7] as well as of their concatenation.[5]

The first model block, the motoneuron, consists of a trans-resistance amplifier (an amplifier with current input and voltage output), high pass filter, and voltage to pulse rate converter. It's output is a series of unit value impulses which exhibit a decline in firing rate (adaptation) with constant current input. The next block, labeled Activation Dynamics consists of a pulse rate to voltage converter, piecewise linear relation, non-linear RC circuit, and multiplier. This block produces a multiplication of the effects of the action potential impulses to simulate the non-linear enhancement found by Burke, et. al.[7]

Finally, the muscle is described by a 2nd order non-linear model obtained by removing a single muscle from the 6th order non-linear full joint model used by Stark and his students.[8,9] Full details of the model equations, the parameter values, and their methods of identification have

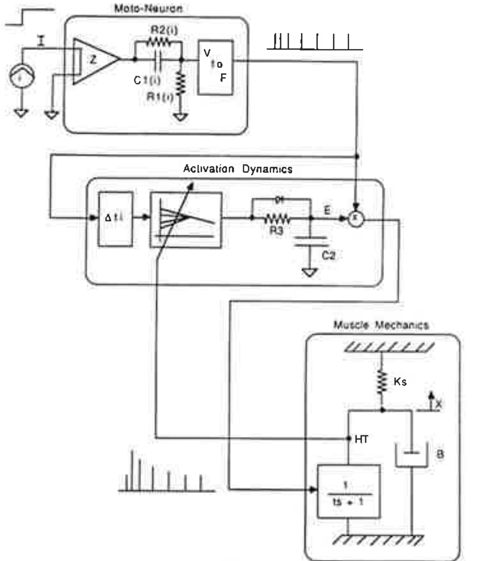

Figure 1.
PEXA model block diagram: model consists of three main blocks corresponding to motoneuron excitation and adaptation dynamics, activation dynamics, and muscle mechanics. Typical signals are plotted at the interfaces.

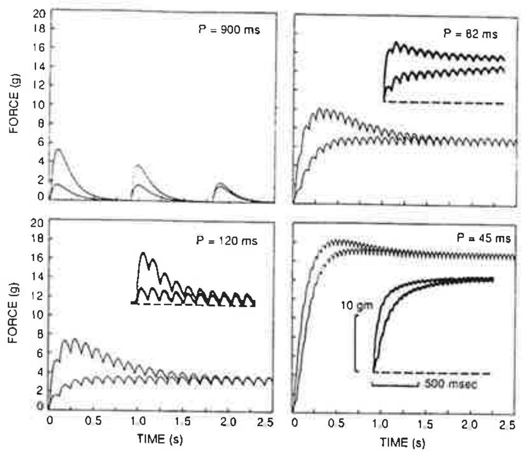

Figure 2.
Model responses to pulse trains. Each graph is the superposition of two responses: force output in response to a step of constant rate pulses, and force output to the same pulse train with an additional pulse 10ms after the first one.

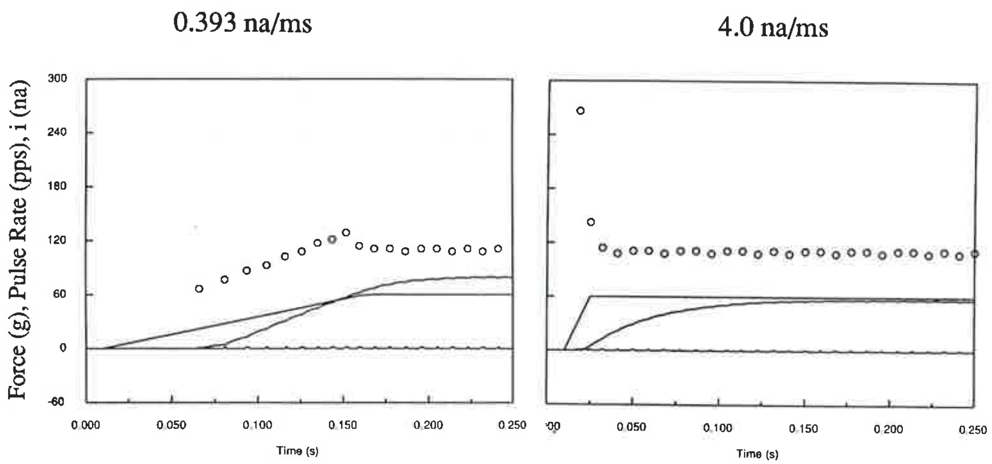

Figure 3.
Response of PEXA model to current ramps of 0.39 and 4.0 na/ms. Shown are current ramps and resulting muscle force (straight and curved solid lines respectively), and instantaneous firing rate of the motoneuron (circles).

been published elsewhere.[1]

The PEXA model was tested and calibrated against experimental data in sections. Two illustrative experiments with the model will be presented here. First, the model will be used to demonstrate the response of the muscle unit to certain pulse trains, second, the model will simulate the response of the muscle unit to ramps of deplarizing current injected into the motoneuron.

Results

The model of the relation of firing rate to muscle force, was tested in a simulation experiment analogous to those performed in-vivo by Burke, et.al.[7] in which force output was recorded in response to pulse train inputs of varying frequencies. For each frequency, two pulse train stimuli were generated. One consisted of a simple train of pulses at that frequency. The second was the same but for the addition of an extra pulse 10 ms. after the first of the train. In each force record (Figure 2, inserts), the higher amplitude signal came from the pulse train input containing the extra pulse, and the "catch like enhancement" persisted over approximately a full second.

The experiment was repeated using the PEXA model for basic pulse train periods of 900, 120, 82, and 45 ms. Force output was recorded in response to pulse train input to the "muscle activation" and muscle mechanics models (Figure 2, main traces). The force responses are comparable to the experimental data[7] which are reproduced as inserts to Figure 2.

The PEXA model can simulate the complete force generation process using a paradigm from Baldissera et. al.[6] who injected ramps of depolarizing current into the motoneuron and observed instantaneous firing rate and tension development. As in the experiments, the PEXA model was driven by current ramps whose slopes ($\frac{di}{dt}$) varied from 4.0 na ms^{-1} to 0.4 na ms^{-1}. The ramps began at t = 0.01 sec, and terminated at a maximum depolarizing current of 60 na. The resulting current, pulse rate, and force outputs (Figure 3) show initial phasic responses in motoneuron firing rate (circles) whose peak rate depends strongly on the current slope and ranges from 308pps at 4.0 na ms^{-1} to 135pps at 0.4 na ms^{-1} with the ocurrence of the maximum firing rate ranging from the second interval at t = .21 ($\frac{di}{dt}$ = 4.0 na ms^{-1}) to the ninth interval at t = .152 ($\frac{di}{dt}$ = 0.4 na ms^{-1}). The force output slope was estimated by fitting a straight line to the force record up to the time of the decline in firing rate to its tonic level (there is no nerve conduction delay in the model). The slopes ranged from 1.11 gf ms^{-1} ($\frac{di}{dt}$ = 4.0 na ms^{-1}) to 0.76 gf ms^{-1} ($\frac{di}{dt}$ = 0.4 na ms^{-1}). The slopes however saturated at about 1.1 gf ms^{-1} for the current slopes above 1.0 na ms^{-1}. Fitting a line to the force slopes below saturation gives a "dynamic gain" of 0.61 gf na^{-1} for the simulated motor unit, a value quite typical of the experimentally measured units.[6]

Discussion

These results confirm an empirical model for the fast dynamics of the generation of muscle forces. Although developed using data from experiments in which steps of current were applied to the motoneuron membrane, it predicts the results of other experiments in which ramps of current were injected. This supports the hypothesis that the non-linear adaptation mechanism is a memoryless function of current level. Phenomena such as the adaptation of moto-neuron firing rate have signifigance for time optimal movements because their time constants are comparable to the duration of EMG bursts in the movement. There are many aspects of the dynamics of very rapid voluntary movements which remain to be understood. For example, the relative amplitudes of the three EMG bursts driving such movements do not match those predicted by earlier simulations[10,11,12] The PEXA model is now ready to be integrated into existing models for the more accurate simulation and understanding of experimental fast movements.

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EMG POWER SPECTRAL ANALYSIS OF ISCHEMIC VS. NON-ISCHEMIC ANTERIOR TEMPORALIS MUSCLE

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INTRODUCTION

A study of the literature on surface electrode EMG suggests that power spectral analysis may be of some significance in the study of fatigued or painful muscles[1]. Authors have suggested that a shift of the EMG power spectrum towards lower frequencies may occur in this fatigued state as a result of decreased action potential conduction velocity[2]. It is hoped that the study of this phenomenon may lead to the development of a diagnostic tool for Myofascial Pain and Dysfunction (MPD) allowing an objective measure of the presence of disease or, as suggested by DeLuca, as a means of evaluating the effectiveness of therapy[3]. Before the study of MPD patients could begin it is necessary to determine our ability to measure shifts in the EMG power spectrum resulting from ischemia and fatigue. The purpose of this study is to reproduce these spectral shifts using a new data collection and fourier analysis program.

METHODS

For this study 23 volunteers were recruited from the faculty, staff, and student body of the University of Maryland Dental School. All subjects were screened to rule out the presence of MPD signs and symptoms by means of a screening questionnaire and clinical examination including muscle and TMJ palpation as well as TMJ auscultation. All subjects were found to be free of the following exclusionary criteria:

Maximum intercuspal opening < 45mm.
Lateral excursion < 7mm.
Protrusion < 6 mm.
Presence of joint sounds.
Sensitivity to palpation in masticatory or associated musculature.

Skin surfaces were prepared for EMG with a skin abrasive. Interelectrode impedance was less than 10K ohms. Surface electrodes were connected to a pre-amp. The signal from the pre-amp was split to drive an oscilloscope for monitoring purposes and an A-D converter for data acquisition and storage with an IBM pc and the Snapshot Storage Scope program of HEM Data Corp. Following data acquisition a fast fourier transform was calculated and the resulting power spectrum was produced using the SFT program of HEM DATA Corp.

Data collection began by recording each subject's blood pressure and then applying a special sphygmomanometer to the scalp over the surface electrodes on the temporalis. A small amount of pressure (30 mm Hg) was applied to the cuff to make certain that electrodes were tightly seated against the skin but not so tight as to impede blood flow. Subjects were then asked to soften a stick of sugarless chewing gum and baseline data was collected while each subject chewed at a comfortable rate. Five one sec. frames of data were collected and saved to disk over a 25 sec. period of time.

Subjects were allowed to relax for approximately 1 minute while the data collection program was reset. The sphygmomanometer was then inflated to the subjects systolic blood pressure plus 30 mm Hg to impede blood flow in the temporalis muscle. Subjects were asked to begin chewing and to signal the examiner when pain was first felt. The onset of pain was used as an indicator for the onset of ischemia and data collection was begun with 5 one sec. frames again being collected while subjects continued chewing for the next 25 sec.

RESULTS

Fast Fourier Transforms were calculated after averaging the five frames of data. Relative positions on the frequency axis were compared between the baseline and ischemic trials by calculating one mean frequency. Figures 1 and 2 depict the Power Spectra of one subject's left side. Note that the mean frequency for the baseline trial is greater than that of the ischemia trial.

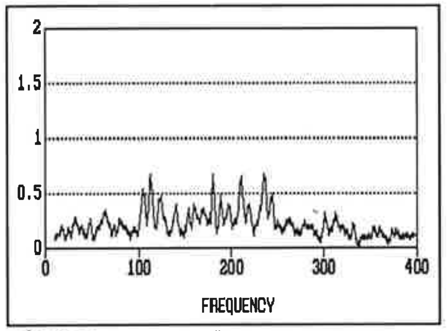

Figure 1 SUBJ # 5 BASELINE
MEAN FREQ = 178

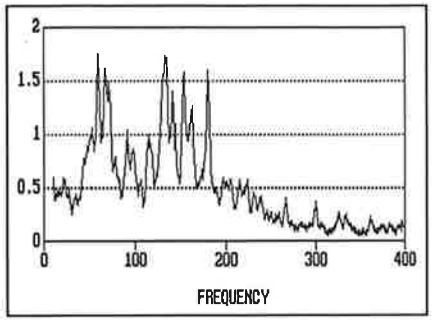

Figure 2 SUBJ. # 5 ISCHEMIA
MEAN FREQ = 158

Basic statistics and a paired samples t-test were calculated comparing the mean frequencies of baseline and ischemic trials for both the left and right side. The results tabulated below show a statistically significant shift of the power spectrum to lower frequencies in the ischemia trials on both the left and right sides.

STATISTICS COMPARING MEAN FREQUENCIES OF THE POWER SPECTRUM

	LEFT BASELINE	LEFT ISCHEMIA	RIGHT BASELINE	RIGHT ISCHEMIA
N OF CASES	23	23	23	23
MIN	113.4	88.6	122.1	93.0
MAX	264.5	241.2	270.3	209.3
STD. DEV.	41.2	38.7	36.3	25.9
MEAN	173.6	144.2	168.4	140.5

PAIRED SAMPLES t-TEST

LEFT		RIGHT
MEAN DIFFERENCE	29.4	27.9
SD DIFFERENCE	19.5	26.3
t = 7.236 DF = 22	P<.001	t = 5.089 DF = 22 P<.001
SIGNIFICANT		SIGNIFICANT

CONCLUSIONS

The data collection and power spectral analysis programs (HEM Data

Corp.) measured a shift in the EMG power spectrum to lower values under ischemic conditions. These results are consistent with those reported previously in the literature(1). This model may prove valuable in determining the physiologic status of the masticatory musculature of TMD patients.

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MISCELLANEOUS

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

ELECTROMYOGRAPHIC ACTIVITY SEQUENCING OF SELECTED MUSCLES INVOLVED IN AN ISOKINETIC DEADLIFT

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INTRODUCTION

Back and torso muscle activity during lifting results not only in muscle tension, but in forces transferred to ligaments and bony structures of the spine. The magnitudes of these forces play a role in low back injury. The observation of powerlifters, who repeatedly lift very heavy loads without a high incidence of injury, suggests that strategies can be adopted to accomplish heavy lifts with minimal potential for injury. The purpose of this study was to compare the activation of major muscles by powerlifters and by asymptomatic control subjects in an isokinetic lift.

METHODS

Four competitive powerlifters and eleven asymptomatic control subjects volunteered for this study. None had suffered a previous injury to the low back and all were in good health.

Each subject had bipolar silver/silver chloride surface type electrodes (NDM Plia-Cell Diagnostic Electrodes, NDM Corp. Dayton, Ohio), with the recording sites set three centimeters apart, installed at the following locations: 1) latissimus dorsi-immediately lateral to the lateral scapular border; 2) erector spinae-1 centimeter lateral to the spinus process of L3; 3) gluteus maximus-on the ischeal spine; and 4) quadriceps-midway between the superior pole of the patella and the bottom of the iliac crest along the anterior thigh.

Isokinetic lifting tests were carried out on a prototype linear lift task machine (Cybex, a Division of Lumex, Ronkonkoma, N.Y.). Following several warm-up and practice lifts, the subject was asked to perform maximal effort lifts during which lifting force, lift height and EMG data were collected at each of two isokinetic test speeds, 30.5 cm/sec or 45.7 cm/sec.

The electrode output signals were amplified by a TECA model TE4 electromyograph and were then passed through a band pass filter with cut offs at 100 and 700 Hz. Each filtered EMG signal was then digitized at 2000 Hz using an Infotek A/D converter model 200 in a microcomputer (Hewlett Packard model 9816). The digitized EMG signals were full wave rectified and smoothed using a 20 Hz low pass filter to create an integrated EMG (IEMG). In order to compare the EMG signals collected from the powerlifters with those of the control subjects, the IEMG data were normalized. This was done using the maximal IEMG value recorded during any of the isokinetic lifts. The lifting force and lift height data were also normalized by their maximal values. Means and standard deviations were determined for the normalized IEMG and the normalized lifting force data as functions of the normalized lift height for each of the two groups.

RESULTS

The force data show that the powerlifters were able to achieve maximal force at 50% of their maximal lift heights, while the control group achieved maximal force at 67% of their maximal lift height. However, only at 33% of the lift height were there statistically significant differences ($p < .05$).

The shape of the normalized IEMG curves were similar between the two groups for the erector spinae, latissimus dorsi and gluteus maximus. The normalized IEMG activity in the quadriceps, however, differed significantly between the two groups. For the asymptomatic control group, the quadriceps activity was maximum at the beginning of the lift and decreased throughout the lift. The quadriceps activity also decreased through the first half of the lift for the powerlifter group, but then increased in the second half of the lift, reaching a maximum at about 83% of the lift height. The differences at 83% of lift height were statistically significant ($p < .05$).

DISCUSSION

The large trunk and leg muscles monitored in this study were chosen because of their role in the lifting process. Other studies have looked at the activity of trunk and abdominal muscle groups (1,2), but no previous studies have looked at the synergistic action of the large trunk and leg muscle groups during a floor to knuckle height lift.

The major difference between the two tested groups appears to be in the activity of the quadriceps mechanism. The quadriceps contraction acts to extend the lower extremity at the knee joint and to stabilize the pelvis (rectus femoris). The asymptomatic control subjects are seen to strongly activate the quadriceps initially, extending the knees. The quadriceps activity then declines to a steady state at 67% to 83% of the lift height. The responsibility for completing the lift is passed to the erector spinae and the gluteus maximus. On the other hand, the powerlifter initiates the lift with combined activity in the quadriceps and the gluteus maximus. The simultaneous extension of both the hips and the knees results in significant early force production, stabilization of the trunk and pelvis and in an improvement of the mechanical advantage of the other muscle groups, particularly the erector spinae. The lift is then completed by knee extension driven by increased activity in the quadriceps, peaking between 67% and 83% of the lift height, and by continued activity of the erector spinae and gluteus maximus extending and stabilizing the hips and spine.

The information from this study may be useful in the rehabilitation setting. First, it is clear that there are a number of muscle groups in addition to the trunk extendors involved in the lifting process. These include those muscles specifically observed in this study, the latissimus dorsi, gluteus maximus, and quadriceps. The rehabilitation of these muscles and muscle groups are just as essential as rehabilitation of trunk flexors and extendors because these muscles provide significant contributions to the total lift effort. Second, the lifting strategies utilized by the powerlifter could be incorporated in work hardening and rehabilitative processes to minimize the load on the erector spinae while distributing it to other trunk muscular components.

ACKNOWLEDGEMENT

This work was supported, in part, by The Akron City Hospital Foundation, The Robertson-Hoyt Fund, and The Department of Biomedical Engineering, The University of Akron.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

INTERRATER RELIABILITY OF VIDEOTAPED OBSERVATIONAL GAIT ANALYSIS

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INTRODUCTION

Gait assessment has become an increasingly important part of an initial and ongoing patient evaluation. Gait assesment is used to determine if the patient's gait differs from "normal," to quantify the degree of abnormality, to identify the causes for the abnormal gait patterns, and as a reassessment tool to evaluate the efficacy of treatment (1,2). Observational gait analysis (OGA) has been the most widely used method of gait analysis. OGA generally consists of joint displacement (kinematic) and/or temporospatial factor analysis.

Recently, videotaping has been used as an adjunct to OGA because videotaping of the patient allows the therapist to view gait patterns repeatedly while avoiding patient fatigue (1,3). Videotaped observational gait analysis (VOGA) also allows the therapist to stop or slow the tape which is thought to increase the accuracy of the assessment (1,3).

The reliability and validity of observations made from videotaping of different populations of patients have not yet been demonstrated. The purpose of this study was to assess the interater reliability of physical therapist's observations made from a videotape of the gait of three rheumatoid arthritic patients.

METHODS

Patients

The three patients videotaped, (ages 43, 45 and 61 years) were previously identified to have abnormal gait pattern secondary to stage 2 or 3 rheumatoid arthritis. Each patient signed a document of informed consent.

Raters

Fifty-four licensed physical therapists employed in seven different hospitals in the Washington, DC metropolitan area were raters for this study. Each rater signed a document of informed consent.

Screening

Initial screening of the patients was carried out by the investigators prior to the study. For reliability measurements to be meaningful, patient variability must be present (4,5). To assure the presence of patient variability and to establish criterion values, gait characteristics of the patients were measured during initial patient evaluation.

Videotaping

Videotaping was carried out using a Panasonic AG HT3 Video Camera and a Bogen 3033 tripod. Each patient was attired in a shirt and shorts and was barefoot. The patients were videotaped from anterior, posterior, and lateral views.

Rating

The raters were oriented to evaluation forms by having all rating categories defined and the scoring protocol explained. While viewing the tape, the raters evaluated the patients' knee joint displacement and gait temporospatial factors. From a lateral view four particular sub-phases of stance were analyzed: initial contact, midstance, heel off, and toe-off. Also analyzed were cadence, stance time, and step/stride length. From an anterior/posterior view analysis of genu valgum and Base of Support was made. Kinematic analysis was recorded on the Knee Joint Displacement Evaluation Form. Comprehensive evaluation of the temporospatial factors was carried out using a Temporospatial Factors Assessment Form.

Data Analysis

Interrater agreement for this study was assessed using the generalized kappa (K) for each of the 10 gait variables across the three patients. The formulae for polychotomous, multirater data (4,6) were used. Agreement among raters for our purposes has been considered equivalent to interrater reliability.

The ICC (6,7) was used to provide a weighting of differences within each variable with the added assumption that the three variable levels were equally spaced (e.g. inadequate = -1, normal =0, excessive = +1). Both ICC (2,1) and ICC (3,1) were computed.

Results

Frequency counts for therapists' ratings of each of the ten variables, generized kappa and ICC's (2,1) and (3,1), are available from the authors. All of the generalized kappas were significantly greater than zero ($z > 1.96$, $p < .01$). However, these kappas were only high enough to indicate slight to moderate agreement (15). The ICC's were significant ($F\ (53,106) > 1.46$, $p < .05$) for the following variables: knee flexion at initial contact, knee flexion at midstance, knee flexion at toe-off cadence and base-of-support.

DISCUSSION

Generally, the agreement coefficients were in the low to moderate range for the ten variables tested. The greatest agreement among the raters for all the variables was in the assessment of genu valgum (k= .52). Raters had more difficulty with the kinematic assessment than with the temporospatial factor assessment. In the kinematic assessment the raters were apparently not sure of what constituted a normal amount of knee flexion for each subphase of stance. They consistently rated patients with greater amounts of flexion than commonly cited in the literature as "normal" (8,9). The therapists did not seem to be familiar with the norms according to the commonly cited references (8,9), and there was a tendency for some of the raters to register an abnormality as being present in the two patients with more apparent joint deformities.

CONCLUSION

Physical therapists demonstrated slight to moderate reliability in measurements of temporospatial and kinematic videotaped gait parameters focusing on the most severly affected knee in three rheumatoid arthritic patients. This study suggests that VOGA has potential as a clinical tool but that there is room for improvement. The authors feel that frequent exposure to VOGA, stressing a more standardized approach to teaching and implementing VOGA, with greater emphasis on the referenced norms, could increase the interrater reliability.

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Electromyographical Kinesiology
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A MODEL FOR INTERPRETATION OF THE FREQUENCY-FORCE RELATION DURING ELECTRICAL STIMULATION OF SKELETAL MUSCLES

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INTRODUCTION

Electrical stimulation of skeletal muscles has been used in physiological studies throughout this century. A decrease in the force response reflects a decrease in the force-generating capacity and changes in time-to-peak tension and relaxation time reflect biochemical events which enable the motor units to keep up tension.

It took until 1977 to add a new measure to this method. Edwards et al then showed that after fatiguing exercise a selective fatigue of long duration occurred at low stimulation frequencies. This phenomenon is called low frequency fatigue (LFF). After this several authors have shown LFF to occur following different types of exercises, both voluntary and electrically stimulated, in humans and animals (Sandercock et al 1985, Alway et al 1987). The significance of LFF is not clear and no evidence has as yet been provided for the proposed mechanisms.

Since this kind of fatigue affects force at low stimulation frequencies more than at high ones, it will have profound effects on the recruitment pattern in voluntary exercise. The muscle must either recruit a larger number of motor units or increase the stimulation frequency in order to keep up the same tension as before. This may be going on for up to 24 hours, thus putting pressure on motor units that would otherwise be at rest.

In our laboratory we have noticed LFF to occur in several studies of voluntary forearm exercise, including low-intensity isometric contractions (Byström & Kilbom 1990).
The aim of this paper is to present a functional model which offers one possible explanation about the origin of LFF.

BACKGROUND

In order to understand the proposed model, the fusion frequencies must be considered. Slow twitch fibers (ST) have time-to-peak tension times in the range of 100 ms or more (Garnett et al 1978) and thus fuse at a stimulation frequency of 10 Hz or more. Fast twitch fibers may have time-to-peak tension times of 40 ms (Eberstein & Goodgold 1968) or even as short as 10 ms (Vander et al 1986) and fuse at frequencies 25 - 100 Hz.

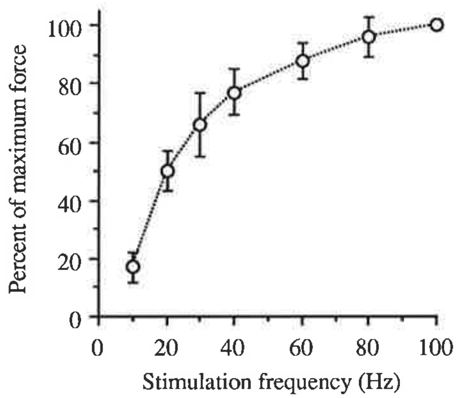

Fig.1
Frequency-force relation in the human forearm extensors during electrical stimulation. 0.7 s tetanus at each frequency. Mean values +-SD for 12 subjects.

As can be seen in fig. 1, stimulation at 20 Hz results in 50-60% of maximum force output (Byström & Kilbom 1990). As the stimulation frequency increases from 20 Hz, the FT gradually fuse from twitches to a tetanus, until maximum force output is reached at 80-100 Hz.

With this background a hypothesis can be formed: At 20 Hz the ST are almost entirely responsible for the force output. The FT contract too at 20 Hz, but their contribution consists of twitches only, and this does not produce much tension.

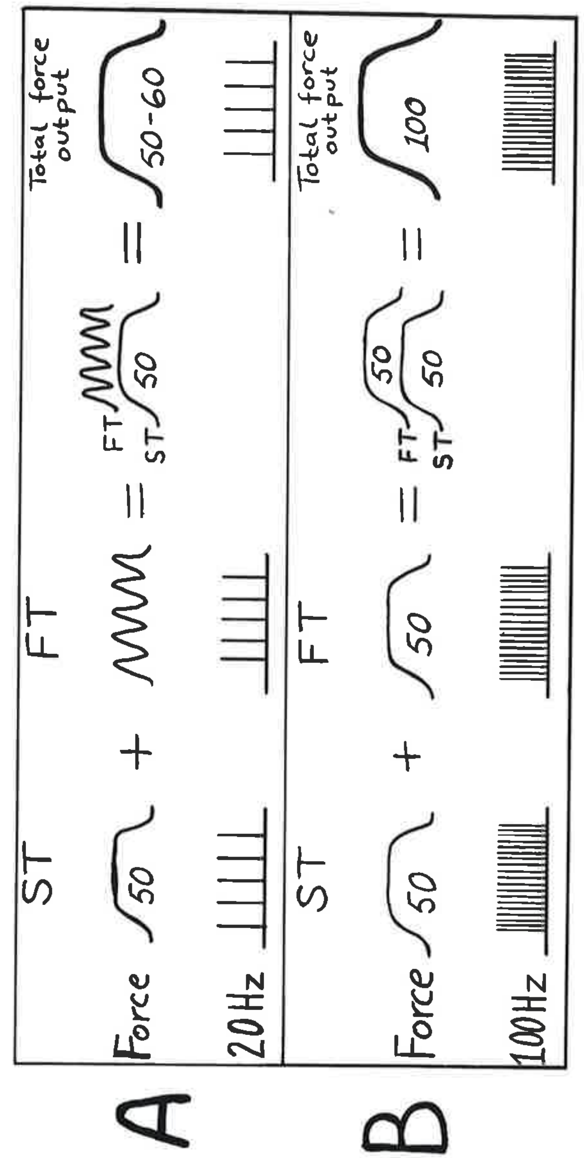

Fig 2.
Theoretical model demonstrating the force response of slow twitch fibres (ST) and fast twitch fibres (FT) at one low (20 Hz, case A) and one high (100 Hz, case B) stimulation frequency. Force in percent of maximal output.

Note that in case A (top) the ST fuse to a tetanus (because of their slow time-to-peak tension times).The FT do not fuse (because of their fast time-to-peak tension times) but contribute to the force output by synchronized twitches. The effective output of these twitches adds, as an example, only about 5-10% to the total force output.

In case B (bottom) all fibres fuse to a tetanus. Hence the ST and FT fibres contribute equally to the total force output. It is assumed that both ST and FT have the same isometric tension per fibre and that they are equally distributed in the muscle.

The model strongly suggests that if the force is reduced at the low frequency more than at the high one, as in low frequency fatigue following i e exhausting contractions, the ST fibres (type 1) are responsible for this force loss.

A SIMPLE MODEL

Assuming the isometric twitch tension per fibre to be equal for ST and FT (Close 1972) and the distribution of ST and FT in the forearm extensors to be about 50-50 (Johnson et al 1973) it may be postulated that if there is a selective loss of force at low frequencies but not at high ones, this must be caused by a force loss in the ST.

The model can be further explained by fig.2 which shows how the two fibre-types contribute to force output when stimulated electrically. Note that the ST constitute a relatively larger part of the force output at 20 Hz (ca 80-90%) than at 100 Hz (ca 50%). Therefore, a selective ST fibre fatigue (drop in the force-generating capacity in the ST) will result in a relatively larger force loss at 20 Hz than at 100 Hz. This is, by definition, low frequency fatigue.

Assuming a selective FT fibre fatigue will lead to high frequency fatigue. Many other assumptions, concerning i e fibre type distribution and fibre type isometric tension, can be tried in the model too. In some extreme combinations the ST fibre fatigue may be less marked. Still, LFF cannot occurr unless there is a drop in the force-generating capacity in the ST that is more pronounced than for the FT.

CONCLUSION

Low frequency fatigue certainly points at a malfunction in the muscles since the force-generating capacity at low stimulation frequencies is reduced. The presented functional model shows that the origin of LFF may be found in the slow twitch fibres (type I).

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

WAVE ANALYSIS OF SURFACE OSCILLATION SIGNALS ON TWITCH CONTRACTION

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INTRODUCTION

Knowledge of muscular action on human is fundamental for improving the treatment of persons with movement disabilities and analyzing the techniques to achieve exceptional performance. Muscle activities can sometimes be recorded directly or can be computed form of electromyographic(EMG) signals. Alternatively, muscular activities can be estimated using models of physical mechanics with kinetic action data. With analyzing at dynamic mechanics, the next step is to determine the muscle components.

The dynamic mechanics of the muscle consists of three components and is shown in Fig.1: elasticity, viscosity and contractibility, and which perform the base function of the muscle contractive action with three elements of muscle length, contractive velocity, and muscle tension. It is difficult to observe about inside of the dynamic components by the non-invasive measurement from outside of the muscle. However the identification of the muscle contractive components has become possible by analyzing complex oscillation waves which has been generated from sources of muscle contraction to the surface of the skin. The purpose of this study was to compare with healthy muscle action and patient's one mechanically. And then this experiment was to find the elastic, viscous and contractile components by the measure of the decrease rate from two measurements of the wave pattern and the frequency of oscillation, and by the amplitude of evoked EMG.

METHOD

Subjects of this study were five healthy men, and five patients of MD disease (Myotonic Dystrophy). Their function were kept walking, and the triceps surae have showed atrophy-involutional phenomena. H-wave on the electrical stimulus was evoked with 4-6mA intensity, M-wave with 18-22mA, and supramaximal M-wave with over 24mA by adding an electric current stimulus percutanously at the knee fossa(poplitea). Furthermore, muscle contraction were practiced with voluntary condition. The surface oscillation which is transmitted to the outside of the skin surface from the inside of the muscle was measured with the Piezo-electric-sensor (manufactured by TEAC). Also, the planter flexible force was recorded by the measure of the charge-amplifier with Washer-type tension sensor (manufactured by Kistler). During muscular contractions, the oscillation waves were super-imposing averaged at 10 times with the trigger of the electric stimulus according to each stimulus condition, and the damping rate and the frequency analysis were calculated by the measure of the decrease rate and FFT(Fast Fourier Transform) used with the analyzer of Nikolei-4094 type computer. The place in which sensor and electrode was installed are shown, and experimental block diagram and a example of oscillation waves are drawn in Fig.2.

O Decrease rate

$$\tan \delta = \Delta / \pi$$

$$\Delta = nT$$

$$n = 1/TT \cdot \log(a_0 / a)$$

a : amplitude of the first wave

a_0 : amplitude of the last wave

T : frequency of the last wave

TT : the decrease time of wave form

O Fourier Transform

$$F(f) = \int_{-\infty}^{+\infty} x(t) e^{-2\pi f t j} dt$$

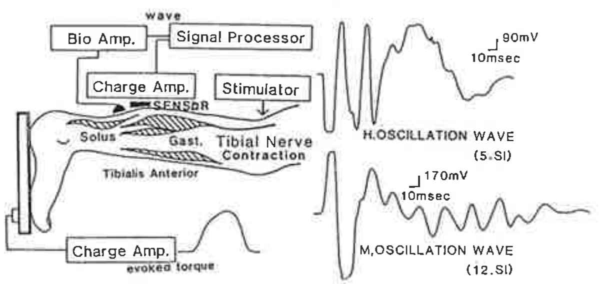

Fig.2 Chart on Tricep surae of the lower leg, and an example of oscillation waves during H-wave and M-wave twitch.

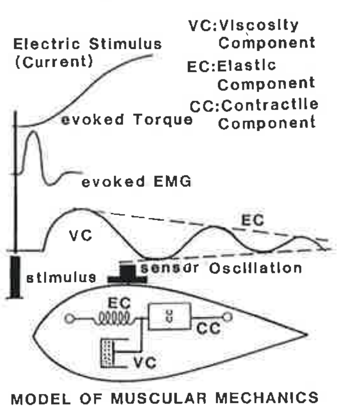

Fig.1 Model of muscle mechanics and oscillation sources of Viscosity, Elastic and contractile components.

RESULTS

Typical power spectra of those surface oscillation are shown during three conditions of H-wave, M-wave, and voluntary muscle contractions about healthy and patient persons in Fig.3. Surface oscillation changed remarkably in proportion to the amplitude of evoked EMG and the intensity of the electric stimulus, and after about 10msec on appearance of evoked EMG, evoked torque was

generated. The amplitude of torque was also generated in proportion to the intensity of the muscle contractive potential. Surface skin oscillation waves also changed in proportional to stiffness and composition of the muscle. In healthy man, there was observed a longer oscillation wave that has damping time of 1,020msec. But averaged damping time of healthy men was 332.8msec. Sustained oscillation was recorded to be notable in the case of MD patients, and their average time was 2,106.0msec. The mean of the decrease rate which are shown at Fig.4, were calculated as follows: 0.20(H-wave), 0.14(M-wave), 0.08(SM-wave), and 0.05(voluntary) in healthy men, and 0.09(H-wave), 0.27(M-wave), 0.05(SM-wave), and 0.19(voluntary) in patients. Those figures showed a quite remarkable difference among subjects of healthy and patients. A result of FFT analysis about the oscillation wave, was shown in Fig.4. The PPF1 (the first peak frequency) appeared at 12Hz in healthy men, and the PPF2 (the second peak frequency) was obtained at the vicinity of 24Hz with high power (density). While in patients, only the PPF2 was shown the lower frequency,

Discussion

Statistical association between surface oscillation and muscular contraction can be improved by the use of Fast Fourier Transform (FFT) that include multivariate prediction of the amplitude and frequency of the muscular composition such as elasticity, viscosity

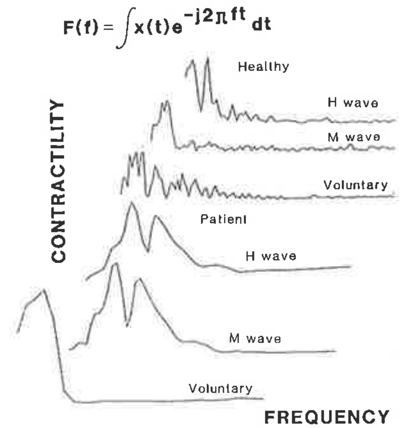

Fig.3 Power spectra of Oscillation wave during H-wave, M-wave twitches with the electrical stimulus.

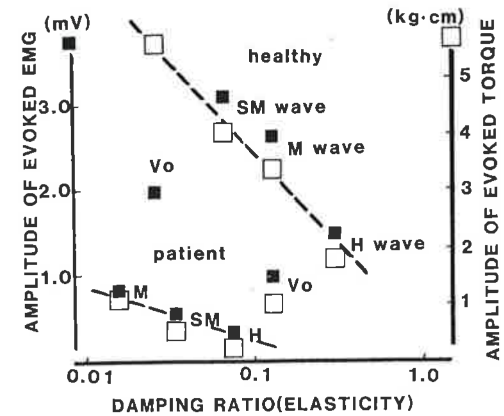

Fig.4 Comparison between healthy and patient with the decrease rate of oscillation waves on the surface skin.

and inertia of the physical muscle model. The damping time would be influenced with subject's central nerve control, which should be reduced to relaxation from the stress, and the antagonistic muscle contraction and inhibitory nerve control had remarkable inferior in the case of patients to healthy men. As a result of analysis,the oscillation waves were patterned a look like sine-wave during twitch contraction. Then, the contractive waves were estimated with the elastic component, viscous component and their contractibility within the muscle physically. The function level of the muscle had influenced on these oscillation waves.

The decrease rate should be depended upon the different function of the elastic component in the muscle, and the sliding velocity of muscular inside filaments clearly occurred as the contractive oscillation of the outside surface. In oscillation waves of the MD patient, tendency of muscle dystrophy would be assumed to be inferior responses of muscular function. This would show the function reduction of the viscous component in the muscle of the patient.

CONCLUSION

In wave analysis on surface oscillation signals during twitch contraction, the change of the muscular mechanics and its functional difference of their components have been possible to be estimated with various parameters of the decrease rate and frequency analysis of FFT, non-invasively. Comparing with both healthy and patient subjects, muscular functional levels were possible to be estimated with frequency analysis of surface skin oscillation of outside signals.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

THE EFFECT OF TENDON VISCOELASTIC STIFFNESS ON THE DYNAMIC PERFORMANCE OF ISOMETRIC MUSCLE

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INTRODUCTION

The dynamic behavior of a muscle-tendon unit is the result of complex interactions between the active and passive mechanical properties of excitable and non-excitable tissues, contractile mechanism dynamics, and load characteristics. Substantial information is available on the dynamic performance of the muscle tendon unit, as well as on the ex-vivo mechanical properties of tendon.

The stress-strain curve of tendon exhibits a toe-off region of low stiffness, which corresponds to the initial straightening of the wavy patterns of fibrils in free tendon. After the collagenous fibrils have been straightened, the stress-strain curve of tendon takes on a stiffer linear characteristic. Further studies of the ex-vivo and in-vivo mechanical properties elucidated changes in the stiffness of tendon as a function of strain rate, and pointed out that tendon exhibits stress relaxation properties associated with viscoelastic materials (Proske & Morgan, 1986; Fung, 1982). Another interesting property of tendon is that long tendons have more compliance than short tendons, allowing the contractile mechanism to shorten more for a given force level. The role of these properties in the dynamic performance of muscle has been more difficult to ascertain, although Proske and Morgan suggested that tendon properties became important during eccentric contraction, where forces much larger than the maximum isometric tension could be achieved.

The objective of this study was to determine the effect of tendon viscoelastic properties upon the resulting dynamic response of a musculo tendon unit during contractions of commonly used force levels and velocity. It was suggested that sinusoidal force variations spanning from 20 to 80% of the maximum isometric force at frequencies from 0.4 to 6 Hz would be representative of most force variations attained during everyday activities except during extremely strenuous activities.

METHODS AND RESULTS

Four adult cats were anesthetized with chloralose (60 mg/kg). Their sciatic nerve was exposed and all branches denervated except that to the tibialis anterior (TA). Two bipolar electrodes were placed on the nerve for late connection to the stimulation system. The T.A. distal insertion was freed and the ankle joint was disarticulated. Pins were inserted through the femoral condyle and mid-tibial shaft to ensure rigid fixation. The animal was rigidly mounted on a platform, and its T.A. tendon was attached through a tendon holding device to a GRASS FT-10 transducer.

The stimulation system has been described and validated elsewhere (Zhou, et al., 1987; Baratta, et al., 1990a,b). Briefly, a computer controlled stimulation system was used to induce the orderly recruitment of motor units with concurrent increase in the firing rate of the active units. Sinusoidal input signals were used to produce sinusoidal force stretch-release cycles, which at 0.4 Hz spanned from 20% to 80% of the muscle's maximal available force at 0.4 Hz. Trials were then performed at .6, .8, 1, 1.2, 1.4, 1.6, 1.8, 2, 2.5, 3, 4, 5, and 6 Hz.

Data was collected by an IBM-AT at 64 Hz, windowed and put through an FFT. A point in the transfer function was defined by the following equation:

$$\frac{\bar{F}}{\bar{V}} = \frac{|F|}{|V|} \angle \Phi$$

F and V are the FFT's of the force and input signal at the trial frequency and Φ is the phase difference between them.

The gain of the data points was normalized with respect to the frequency of 0.4 Hz and converted to dB. Gain and phase were then plotted in conventional Bode gain and phase plots (Fig. 1a). The method of least squares was used to obtain the best fit pole locations for a critically damped second order linear model (Baratta et al., 1990; Baratta & Solomonow, 1990). This process was repeated after shortening the tendon by 2 cm (Fig. 1b).

The pole locations obtained with long tendon were compared with those obtained with the shortened tendon by using paired T-tests. These tests showed no statistically significant difference between the pole locations obtained with short and long tendon. Harmonic distortion calculations revealed no significant influence in linearity.

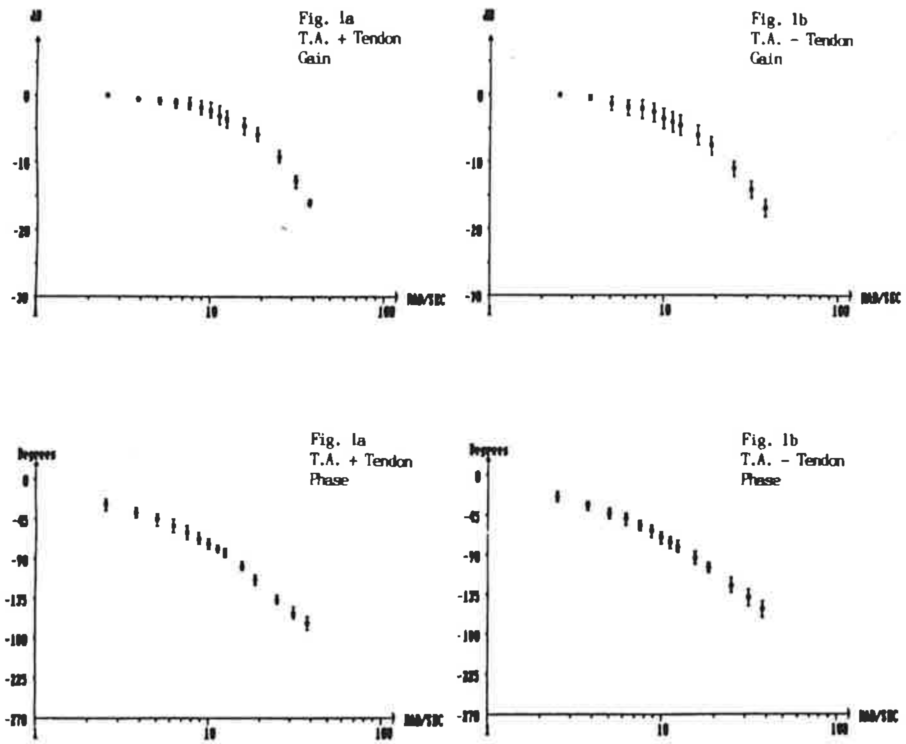

Fig. 1 - Gain and phase plots for T.A. muscle before (1a) and after (1b) shortening the tendon. Note the similarity between the plots, indicating no significant influence on the overall dynamic response of this muscle.

DISCUSSION

The most important fact emerging from this study is that during isometric contractions ranging from 20% to 80% of the muscles maximum isometric tension, the tendon viscoelasticity has no significant effect upon the dynamic response of the muscle tendon unit. Under these conditions, the tendon acts as rigid force transmission mechanism.

The effect of non linear tendon stress-strain curve had manifested itself in previous studies (Zhou et al., 1987) when a toe-off was observed in the linear 20% of the force generation cycle. In the mid range, however, such non-linearity was not evident, and tendon length was not observed to be a significant factor in pervious multifactorial analyses (Baratta & Solomonow, 1990).

The oscillations in force as result of the input stimulus variations indicate that throughout the cycles, muscle fibers must have shortened and elongated to allow the mi crofibrils to generate force. Since the overall muscle-tendon unit did not change, and since tendon did not stretch significantly, one must conclude that the largest amount of stretch occurred in the aponeurosis, as predicted by Huijing and Ettema (1988, 1989). The aponeurosis was included in both experimental conditions, so any changes owing to this structure would not be revealed by this experimental technique.

In summary, the results of this study suggest that during isometric contractions spanning from 20 to 80% of a muscle's maximum isometric force, the viscoelastic properties of tendon do not alter the muscles performance and allow fast, precise motion by serving as a stiff force transmission linkage.

ACKNOWLEDGMENT: This work is supported by NSF Grants EET-8613807 and EET-8820772.

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Electromyographical Kinesiology
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MECHANICAL MODULATION OF BRAKING TORQUE IN RAPID MOVEMENTS

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INTRODUCTION

Rapid movements about a single joint are generally believed to be controlled by a triphasic pattern of EMG activity (1,2), yet only the role of the first of the three bursts is well characterized. In movements made over a range of amplitudes, investigators have been puzzled as to what role the antagonist plays in the deceleration of the movement. Karst and Hasan, 1987 (3) found that antagonist activity varies according to the requirements for braking. However, with the instructions to move "as fast as possible" others have found that antagonist activity is unrelated to movement amplitude (4,5,6). In "fast as possible" wrist flexions, we found that braking torque scales precisely with accelerating torque (and agonist EMG), which in turn scales with movement amplitude. We did not, however, find corresponding scaling of anatagonist EMG activity. We tested the hypothesis that scaling of braking torques is mediated simply by the mechanics of the stretching antagonist muscles. Such mechanical mediation might occur either as a result of the force-velocity properties of the lengthening antagonist muscle, or by an appropriate change in energy storage in its series elastic element. To evaluate the former mechanism, we imposed graded velocities of stretch on partially activated wrist muscles.

METHODS

Task Description

Subjects were seated at a table with the right hand placed in a manipulandum allowing wrist movements in the horizontal plane only. An oscilloscope placed directly in front of them displayed a position cursor. Subjects moved their position cursor into a narrow target centered on the screen (neither a flexed nor extended position) and maintained this position against a range of preloads from 0.0 to 6.0 Nm. After a random delay, a motor, attached to the handle, was used as a velocity servo, imposing an additional torque pattern to achieve a constant velocity of wrist rotation. Subjects were instructed "not to voluntarily intervene" either to oppose or to assist the ramp stretch.

<u>Position ramp inputs</u>

Subject's hands were moved at velocities reaching 150 to 600 degrees per second, in increments of 75 degrees per second. The device used to produce preloads and position ramps was a permanent magnet motor (Electrocraft Series 720), operated as a velocity servo, digitally controlled by an IBM PC/AT. The motor was driven by the sum of a constant input (the preload) and a multiple of a velocity error signal--the difference between a command velocity set by the experimenter and an estimate of motor velocity, obtained by differentiating the position signal. The hand reached the command velocity after 10 msec. The velocity profile during the rise to attained command velocity was independent of preload.

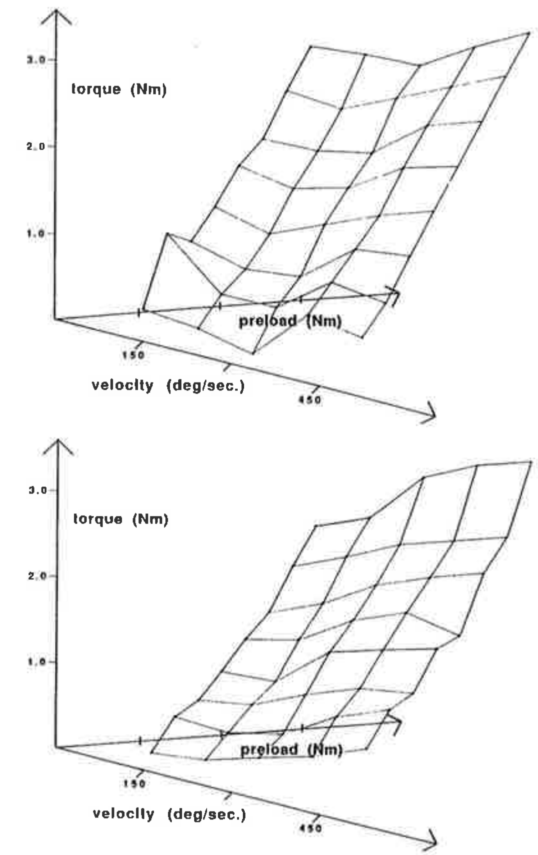
<u>Data Collection and Analysis</u>

Position, torque, acceleration and two surface EMG's (flexor carpi radialis and extensor carpi ulnaris) were sampled 500 times per second. Because the torque records contained large inertial components, in addition to the muscle-produced torques we wished to measure, we subtracted torque records from trials without preload from those from preloaded trials with the same velocity. We verified that this procedure correctly identifies the muscle-produced torque by testing on an artificial subject (mass for hand, spring for muscle, load cell to measure muscle force).

RESULTS AND DISCUSSION

Figures 1a and b display the peak reduced torque (total torque-inertial torque) produced in response to variation of initial preload and stretch velocity for two subjects. The three dimensional plots show that most of the increase in torque production is due to the increase in preload. We found increased stretching velocity to have almost no influence on torque production.

Fig. 1a and 1b. Reduced torque produced in response to variation of stretch velocity and initial preload.



Braking torque is precisely scaled to driving torque in the movements we measured. Scaling by active state is unlikely, however, given the lack of modulation of antagonist EMG. Our results, and those of others (7) do not support scaling by stretch velocity.

The remaining mechanical element of the stretched muscle, the series elasticity, may mediate the observed scaling. Energy is no doubt stored in the antagonist's series elasticity, as the antagonist is actively stretched (after arrow in Figure 2). Larger movement amplitudes would produce larger stretches of the series elasticity, and correspondingly larger braking forces.

The movement is near its maximum excursion when peak decelerating torque is achieved (Figure 2), so hand velocity is small. If the contractile element is also nearly isometric at this time, then the series spring lengthens in proportion to movement amplitude, and produces decelerating forces according to its length.

Simulation of an identified model of wrist dynamics (8) supports this explanation (Figure 3). Peak deceleration occurs at slow stretch velocities and appropriate scaling of peak antagonist torque is not strongly dependent on the form or slope of the assumed lengthening force-velocity curve.

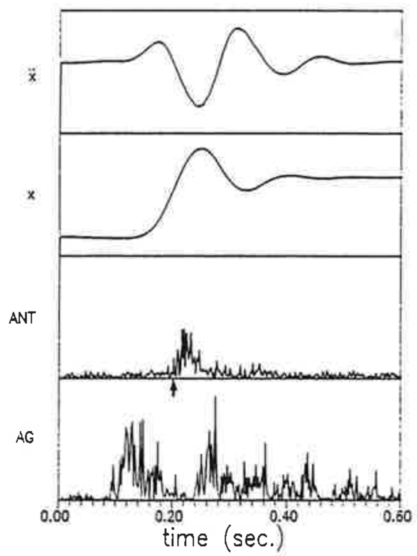


Fig. 2. Acceleration, position, antagonist, and agonist EMG traces recorded during a typical rapid movement.

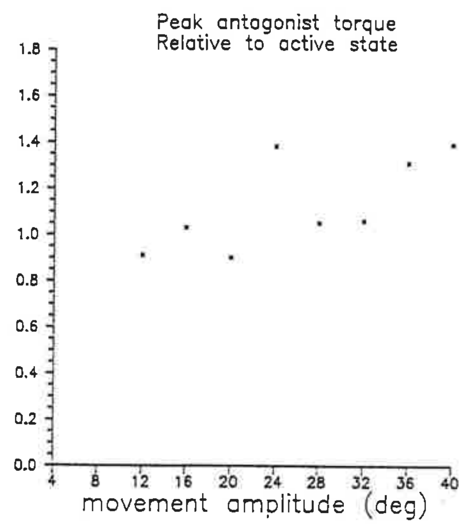


Fig. 3. Peak antagonist torque does not exceed active state over the full range of movement studied.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

MAGNITUDE DIFFERENCES DURING THRESHOLD EXCITATION OF FOREARM AND LEG PERIPHERAL NERVES

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INTRODUCTION

Present clinical uses of transcutaneous electrical stimulation is frequently associated with the excitation of sensory and motor nerves. Commercial stimulator's output includes many different waveforms, varying pulse parameters and numerous clinical problems that require stimulation of both the upper and lower extremities.[1] Several published studies have attempted to identify a preferred waveform during neuromuscular excitation.[2-7] Most of these studies tested the responses to different waveforms in the lower extremity at maximally tolerated excitation levels or at fixed levels of muscle torque production. No study could be found where both upper and lower extremities were stimulated and the threshold, rather than maximum or moderate excitation of sensory and motor fibers, was established. Recently our group has completed the first study in which five of the more clinically common waveforms were used to induce threshold excitation of sensory and motor nerves in the forearm and leg.[8] As a spin-off data from the aformationed study we noted a clear differences between the upper and lower extremities. Therefore, the purpose of this presentation was to compare magnitude differences during threshold excitation of forearm and leg peripheral nerves.

METHOD

Eighteen healthy subjects with a mean age of 26.9 years volunteered to participate in the study which was approved by the IRB of the University of Maryland and the CDRH IRB. Each subject signed a consent form after which she/he assumed a sitting position with their right lower extremity resting on a chair while the knee was in full extension and the foot supported in zero degrees dorsiflexion. The right upper extremity was positioned on an arm rest with the shoulder in the anatomical position, the elbow at 90 degrees flexion, the forearm in full pronation and the hand and fingers fully supported yet relaxed on the arm rest.

The stimulation system consisted of two constant voltage signal generators (WAVETEK model 175 and HP model 3314A) controlled by an HP 5000 Microcomputer computer and an HP model 5180A waveform recorder. Using a special software the system generated five different waveforms.(Fig.1) The phase duration was 200 Usec and repetition rate of 50 Hz. Each of the five waveforms was delivered in sequential order via two self-adhesive rectangular electrodes, each 8.8 cm long and 3.8 cm wide.

The right forearm was wiped clean with tap water and the proximal electrode was placed perpendicular to the segment on the dorsal surface over the extensors muscle group, 4 cm distal to the lateral epicondyle. The second electrode was placed 8 cm distal to the first electrode. During stimulation of the lower extremity the same electrode application procedure was repeated over the dorsiflexor muscle group, but the placement of the proximal electrode was 10 cm distal to the knee joint space.

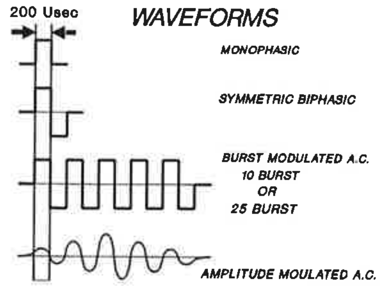
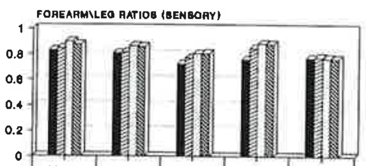


Fig. 1. Waveforms illustration

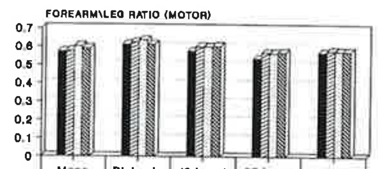
RESULTS

Peak voltage, peak current, phase charge, and total charge were recorded during threshold sensory or motor excitation of the forearm or leg. The data obtained during threshold sensory excitation of the forearm were divided by the corresponding data of the leg to yield relative, normalized values of sensory threshold. The same procedure was used to calculate relative motor threshold excitation. The data is summarized in Fig.2 .



	Mono	Biphasic	10 burst	25 burst	AM modul
PEAK CURRENT	0.829	0.806	0.723	0.759	0.77
PEAK VOLTAGE	0.845	0.815	0.769	0.846	0.775
PHASE CHARGE	0.898	0.863	0.802	0.881	0.768
TOTAL CHARGE	0.875	0.857	0.806	0.878	0.764

PEAK CURRENT · PEAK VOLTAGE · PHASE CHARGE · TOTAL CHARGE



	Mono	Biphasic	10 burst	25 burst	AM modul
PEAK CURRENT	0.572	0.613	0.576	0.636	0.57
PEAK VOLTAGE	0.577	0.625	0.6	0.568	0.578
PHASE CHARGE	0.603	0.635	0.599	0.566	0.575
TOTAL CHARGE	0.59	0.612	0.603	0.57	0.575

PEAK CURRENT · PEAK VOLTAGE · PHASE CHARGE · TOTAL CHARGE

Fig. 2. Ratios of forearm\leg threshold sensory and motor excitaion

Each measured variable were analyzed by ANOVA with the waveform and sensory\motor excitation being the two main factors. Results indicated significant differences between sensory and motor excitation of all measured variables irrespective of waveform. The means for each of the measured variables are presented in Table I.

TABLE I

MEANS OF MEASURED VARIABLES

Measured Variable	Segment	Sensory	Motor
Peak Current (mA)	Forearm	30.90	55.00
	Leg	39.70	95.70
Peak Voltage (V)	Forearm	3.44	6.35
	Leg	4.27	10.81
Phase Charge (Uc)	Forearm	1.13	2.03
	Leg	1.35	3.43
Total Charge (Uc)	Forearm	27.11	48.90
	Leg	33.35	84.58

The average phase charge of all waveforms during sensory excitation of the leg was 15.7 percent higher than that required to excite the forearm sensory fibers. Threshold excitation of motor nerves was induced with 40.4 percent higher phase charge in the leg compared to the forearm. Very similar results were recorded with the other three measured variables.(Fig.3) Combining the ratios of peak voltage, peak current, phase charge, and total charge of all five waveforms resulted in the respective sensory and motor excitation thresholds of 18.4 ± 4.0 and 41.3 ± 2.3 percent higher in the leg compared to the forearm.

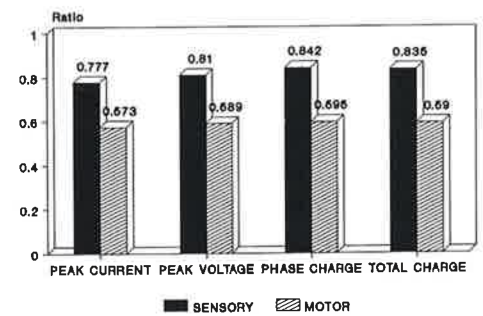


Fig. 3. Summary of ratio's for each of the measured variables

CONCLUSION

Transcutaneous stimulation of sensory and motor nerves required significantly higher magnitude of selected pulse parameters during stimulation of the leg compared to the forearm. Motor excitation required significantly higher magnitude of stimulation than sensory excitation. The source of the demonstrated magnitude difference is at present unclear.

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Electromyographical Kinesiology
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EMG-VELOCITY AND FORCE-VELOCITY CHARACTERISTICS OF HUMAN TRICEPS SURAE MUSCLE: EXPERIENCES FROM IN-VIVO TENDON FORCE MEASUREMENTS DURING NORMAL RUNNING

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INTRODUCTION

Measurements of the mechanical responses during isolated concentric and eccentric actions have produced relatively similar force-velocity curves in animal (in situ) preparations (e.g. Hill[4]) and human experiments with voluntary activation (e.g. Wilkie[9]; Komi[5]). Natural human and animal locomotion are, however, different from these isolated muscle actions. Normal muscle-tendon function is characterized by repeated stretch-shortening cycles (SSC), were eccentric (ECC) and concentric (CONC) actions follow in sequence (e.g. Komi[7]). Stretching of an active muscle-tendon unit prior to its shortening modifies the mechanical (e.g. Cavagna et al.[1]), and electromyographic (EMG) responses (e.g. Komi[6]) of the CONC action. The purpose of this paper is to present evidence that during the functional phase of the natural SSC both the mechanical and EMG curves of the ECC and CONC action represents responses different from those measured during isolated muscle actions. The present study utilized direct in-vivo force measurements of the human Achilles tendon (AT) during running at different speeds.

METHODS

Two male subjects participated in this study. They were 35 and 41 years of age at the time of the experiment and both were in excellent physical condition. They were aware of potential risks involved in the surgical implant and testing procedures prior to their participation and each gave their separate written consent. Strict guidelines were followed regarding the use of human subjects in these experiments.

A buckle-type force transducer was surgically implanted under local anesthesia on the right AT. Details of the transducer construction, calibration and operation have been previously reported in detail (Komi et al.[8]). The surgery lasted 15-20 minutes, and was followed by a calibration procedure (10 minutes). The actual measurements lasted approximately two hours, after which the transducer was removed.

Each measurement included several activities, e.g. jumping, running, hopping and cycling. The running was performed over a long force platform using various constant speeds. The subject carried around his waist a telemetric system

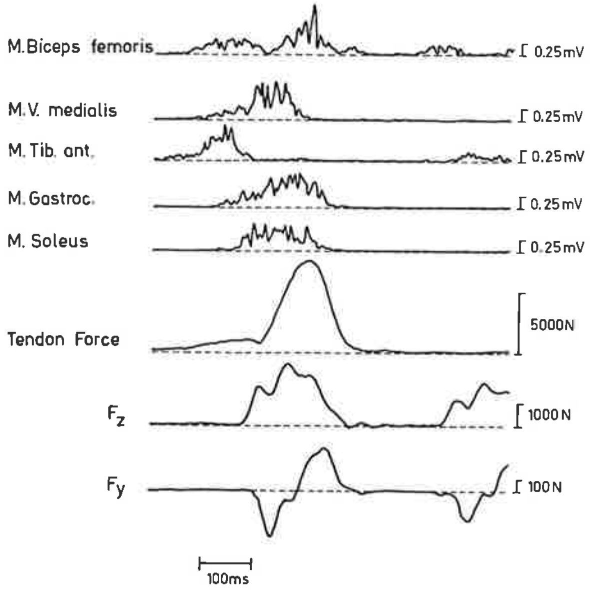

Fig. 1. A representative record of Achilles tendon force response together with F_z and F_y ground reaction forces and selected EMG activities when the subject was running at 5.6 $m \cdot s^{-1}$.

(Medinik) containing the transmitter units for EMG and AT force (ATF). Bi-polar surface electrodes (Beckman miniature, 4 mm diameter) were placed on the soleus (SOL) and gastrocnemius (GAST) muscles. Each performance was filmed from the side view with a Locam camera set to operate at 100 frames$\cdot s^{-1}$. The film analysis was used to estimate the percent change of the length of the GAST and SOL muscles from the angles of the knee and ankle joints according to the method reported by Grieve et al.[3].

RESULTS AND DISCUSSION

An example of ATF response during foot contact together with F_z and F_y ground reaction forces and EMG activities of the selected muscle groups are presented in Figure 1. The peak ATF occurred approximately at the time when the F_y force component became positive. Applying the technique of estimating length changes of the total muscle-tendon complex, the force length (F-L) and force-velocity (F-V) curves (Figure 2) could be obtained during the period of ground contact. This represents the phase, where the functional SSC can be identified (see Komi[7]). The shapes of the curves in these exemplar figures support the position that force

can be enhanced during the CONC phase. When running was performed at medium and high velocities, the total loop in the F-L curve was smaller at the higher speed. The peak force appeared lower while the rate of the force development during the ECC part was very high and occurred with a small change in length of the muscle-tendon complex. F-V curves were then calculated from F-L data. As shown on the right side of Figure 2 the form of this curve is very different from the classical F-V-relationships. EMG-length and EMG-velocity curves for the SOL muscle are presented in Figure 3 and illustrate that the activation of the SOL muscle increased during the first half of the stretching (ECC) phase, decreased toward the end of this phase and remained at low levels during the entire CONC action. For the GAST muscle the EMG responses were, in principle, similar.

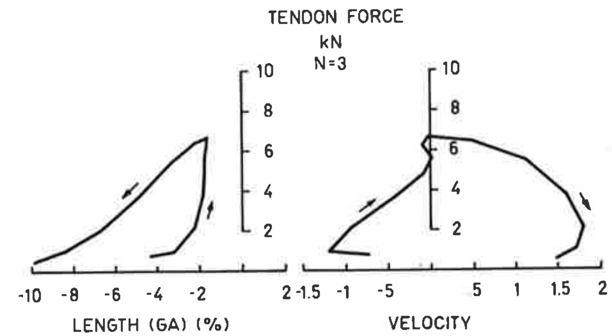

Fig. 2. Force/length and force/velocity curves of the gastrocnemius muscle during the ground contact phase when the subject ran at very high velocity (9.0 $m \cdot s^{-1}$).

The present results of the F-L and F-V curves are in agreement with the findings of Gregor et al.[2], who measured mechanical outputs of the cat soleus during treadmill running. They demonstrated further that the force generated at a given shortening velocity during late stance was greater than the output generated at the same shortening velocity, in situ. Although the present study could not utilize the classical F-V measurements, the form of the curves clearly indicates force enhancement in the CONC part of the SSC. It could be appealing to explain the difference of the F-V curve from that of the classical curve in isolated muscle preparations or in human experiments by differences in the activation levels between the two conditions. When F-V curves are measured in isolation the activation is kept maximal and constant throughout the different velocity conditions. The very low EMG activity in the CONC phase of the present experiment emphasizes that the potentiation of the F-V curve in the CONC phase of the cycle cannot be due to increased EMG. Natural SSC involves controlled release from high forces, caused primarily by the ECC action. This high force favors storage of elastic strain energy in the muscle-tendon complex. Thus the potentiation of the F-V curve during the CONC phase of the SSC is likely due to the fact that a great portion of the stored energy can be recovered during the subsequent shortening phase. It is concluded further that natural locomotion with primarily SSC muscle action may produce muscle outputs, which can be very different from the various conditions of isolated preparation, where activation levels are held constant and storage of strain energy limited.

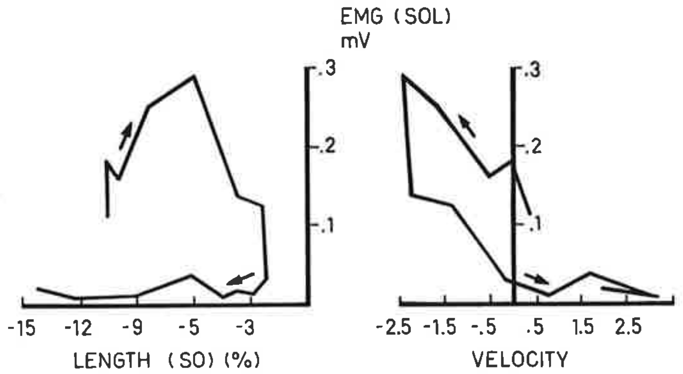

Fig. 3. IEMG/length and IEMG/velocity curves of the soleus muscle for the ground contact phase at a slow speed of running (4.5 m·s^{-1}). The subject was different from that of Figure 2.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

THE EFFECTS OF TENS THERAPY ON MANDIBULAR PROPRIOCEPTION

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INTRODUCTION

The use of Transcutaneous Electrical Nerve Stimulation (TENS) for the relief of chronic oral-facial pain is increasing. The current concept of TENS began with the presentation of the "Gate Control" Theory of pain transmission by Melzack and Wall in 1965 (1). This theory opened new doors, and in 1965, Wall and Sweet used electrical current applied to the skin in an attempt to stimulate large "A" fibers which might inhibit pain transmission by the smaller "C" fibers (2). The results were encouraging and, hence, arose the concept of TENS for relief of pain (2).

Although TENS is known to effectively relieve pain, the mechanism of action is not clearly understood and varies with pulse width, frequency, and intensity (3,4). Three main theories of TENS effect on pain include the "Gate Control" Theory (1), the "Endogenous Opiate" Theory (5), and the Nerve "Blocking" Hypothesis (6,7).

According to a review of the use of TENS in dentistry published in 1986 and a symposium on electronic dental anesthesia held in 1987, most of the current TENS papers are based on individual opinions and observations (8). Because the majority of literature on TENS is retrospective in nature, particularly as it deals with treatment of chronic facial pain, it was felt that a prospective controlled study examining possible treatment side effects was indicated.

In the present experiment, we sought to investigate whether or not TENS applied to the anterior temporalis and superficial masseter muscles adversely affects mandibular proprioceptive ability.

SUBJECTS

Twenty-five subjects, 14 males and 11 females, volunteered as participants in this study. The subjects ranged from 17 to 50 years in age. All subjects had a complete natural dentition with orthodontically acceptable overjet and overbite (1–3 mm). Subjects also had a negative history of parafunctional habits, temporomandibular dysfunction (TMD) or other pathological processes which might affect accurate data collection.

Experimental procedures (discussed below) were explained to the subjects, and written consent was then obtained. Before an experimental run, subjects demonstrated the ability to perform all functional exercises or movements which were to be examined. In each case the subject served as their own control.

APPARATUS

The TENS unit used in this study was the autoTENS, produced by Physio Tronics Corp., 7309 E. 21st Street, Wichita, KS 67206. Self-adhesive electrode pads (3.5 x 2.5 cm) were used in a bipolar method of stimulation, with 1 electrode placed on both right and left anterior temporalis and superficial masseter muscles for a total of 4 electrodes. A series of acrylic discs prepared by the principal author were used to test "Interdental Dimension Discrimination" (9). The discs were of uniform diameter (25 mm), but the thickness varied in 1 mm increments from 3 mm to 12 mm. To test reproducibility of mandibular position, a Sirognathograph and BioPak software package (BioResearch, Milwaukee, WI) were used.

PROCEDURE

Subjects were tested individually during one experimental session lasting approximately 45 min. Two tests were performed both with and without TENS. In the first test, a reference sized disc (8 mm) was used. The subject was asked to incise gently on the reference disc and memorize the thickness. This was done with the subject's eyes closed to eliminate any visual input. The reference disc was then randomly replaced with one of the other test discs, and the subject asked to determine whether the test disc was thicker, thinner, or identical to the reference disc. The reference disc was always replaced between the incisors prior to the trial of a new test disc. The response was recorded as either correct or incorrect. The least noticeable difference (LND) was the range of incorrect answers (mm). This series of trials was performed randomly with both the TENS unit turned on and with the unit off.

The second test involved three parts: vertical, protrusive, and right lateral reproducibility of mandibular movement. With the TENS unit still attached to the subject, a Sirognathograph was then placed on the subject's head. A small magnet was then adhered to the subject's mandibular incisors using stomadhesive paste. The sequence for testing with TENS-on and TENS-off for each subject was randomly determined. The subject was then asked to begin moving his/her mandible in 1 of the 3 previously mentioned directions.

The subject was then asked to stop at a predetermined point. This point was recorded by the author and memorized by the subject. The subject then closed to the original starting position and attempted to return to the first position. The error was recorded as the difference in the 2 readings. Five trials were done in all 3 directions with and without TENS.

The parameters used with the TENS were high frequency (90 Hz.), a pulse width of 50-350 microsec (100% modulated 2 x 1.2 secs), and an intensity set just below that which produced visual muscle contraction.

RESULTS

One tailed t-tests were run to analyze the results of the 2 conditions (with and without TENS) for the 4 test situations (LND, vertical, protrusive, and lateral reproducibility of jaw position). However, no significant differences were found for any of the 4 situations at a $P < .05$ level of confidence. Table I summarizes the group means, median values, standard deviations, and standard errors.

TABLE I

GROUP MEANS, MEDIANS, STANDARD DEVIATIONS, AND STANDARD ERRORS

	LND W/O TENS	LND W/TENS	Vert. W/O TENS	Vert. W/TENS	Protr. W/O TENS	Protr. W/TENS	Rt. Lat. W/O TENS	Rt. Lat W/TENS
N	25	25	25	25	25	25	25	25
MEAN	2.48	2.40	2.10	2.38	0.97	1.13	1.43	1.32
MED	3.00	2.00	1.88	2.02	0.84	1.10	1.32	1.20
SDEV	0.87	0.87	1.06	1.23	0.43	0.40	0.64	0.44
SE	0.17	0.17	0.21	0.24	0.08	0.08	0.12	0.09

In conclusion, this study found no significant difference in mandibular proprioceptive ability when TENS is applied to the superficial masseter and anterior temporalis muscles. However, this study dealt only with one aspect of proprioception, muscle length. Future studies might be extended to other aspects of proprioceptive, such as muscle tension.

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NEUROMUSCULAR DISEASE

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

THE EFFECT OF PULSED MAGNETIC STIMULATION ON LEARNING BY PASSIVE AVOIDANCE METHOD

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INTRODUCTION

The magnetic stimulation with short duration is able to excite the nerves and it enable us to study the nerve conduction in central nervous system [1)2)3)]. Although no deleterious effects have been observed thus far, the safety of this technique has not fully been established [1)2)]. Our recent study [4)5)] showed the decreased dopamine and serotonine were observed 60 minutes after 50 times pulsed magnetic stimulation. These results tempt us further investigation of the effect of the pulsed magnetic stimulation of the brain with regards to the learning function by using the passive avoidance method.

MATERIAL AND METHODS

The detail of our pulsed magnetic discharge system was reported in others [3)4)6)7)]. Its magnetic field was 2.34 Tesla with an inductance 7 μH and a peak time 137 μs. Mice were placed in the long thin circular chamber with cover by meshes around their head, and received the pulsed magnetic stimulation (MS), 50 times or 10 times in 0.1 Hz by a flat circular coil, or electroconvulsive shocks (ECS).

Two hundred twentysix normal slc : ddY male mice each weighing 27-29 g were used in this study. The study was planned in 4 sessions in different mice. The mice were separated into three group in each session.

In the first session : Mice in MS group received MS, 10 times, one hour before the learning in the passive avoidance method. Mice in ECS group received ECS one hour before the learning. In the second session : Mice in MS group received MS, 10 times, one hour after the learning. Mice in ECS group received ECS one hour after the learning. In the third session : Mice in MS group received MS, 50 times, one hour before the learning. Mice in ECS group received ECS one hour before the learning. In the fourth session : mice in MS group received MS, 50 times, one hour after the learning. Mice in ECS group received ECS one hour after the learning. In each session mice in the control group received neither MS nor ECS before or after learning.

	number of mice	learning trial	test trial	number of mice with over 300 sec of latency in test trial
First Session				
MS	20	15.8 ± 6.3	110.7 ± 71.0	1
ECS	20	16.6 ± 8.6	27.2 ± 16.7**	0
Control	20	13.4 ± 6.3	124.3 ± 91.2	2
Second Session				
MS	15	26.6 ± 8.4	191.7 ± 89.6	4
ECS	12	24.5 ± 13.0	55.5 ± 42.3**	0
Control	15	24.1 ± 11.1	176.2 ± 108.1	6
Third Session				
MS	23	22.6 ± 11.8	167.1 ± 106.8	5
ECS	18	19.9 ± 6.2	57.9 ± 44.9**	0
Control	20	20.8 ± 7.6	116.1 ± 82.5	2
Fouth Session				
MS	20	14.7 ± 5.2	150.7 ± 99.5	5
ECS	23	14.1 ± 5.6	49.6 ± 65.7**	0
Control	20	14.5 ± 6.0	143.3 ± 105.9	5

** : Kruskal-Wallis H-test, χ^2 -test ; p<0.01

Table 1 : Time (sec) of latency (mean ± 1 standard deviation) in learning trial and test trial.
MS : magnetic stimulation, ECS : electroconvulsive shock.

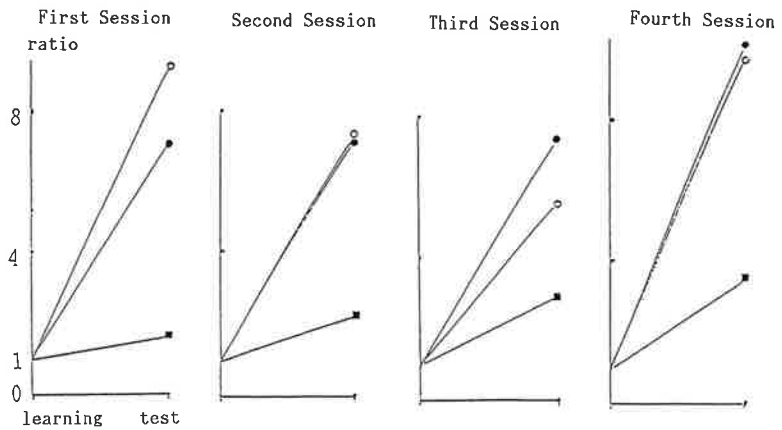

Figure 1 : The ratio of latency of test to latency of learning in passive avoidance method.

- • magnetic stimulation group, o control group,
- ■ electroconvulsive shock group.

Apparatus of passive avoidance

The passive avoidance equipment [8)9)10)] consisted of two compartments (light and dark chambers). The light chamber was constructed entirely of clear plastic and had inside dimensions of 7 ×9 ×14 cm. It was illuminated by a 15 W incandescent light. The dark chamber had a wooden roof and sides and the interior was painted black (inside dimensions- 14×14×14 cm). The floor consisted of stainless steel bars which were 3 mm in diameter and placed 7 mm apart. A foot shoch of constant current DC pulses could be delivered to the bars by an shock generator and scrambler (type 11-13, BRS/LVE company). The light and dark compartments were connected by a 3×3 cm hole and this could be covered by a sliding metal door.

Procedure of passive avoidance [10)]

In day 1 the learning procedure was to place the mice in the light chamber so that it faced away from the hole leading into the dark chamber. The animal's latency to step through into the dark chamber was then measured with a stop-watch. The end-point taken was when the animal had all four paws on the grid floor of the dark chamber. The mice were tested (day 2), 24 hours later, using the same procedure except that no shock was delivered. If the mice did not step through into the dark compartment within 300 sec, the trial was terminated and the latency to step through was recorded as 300 sec [9)].

RESULTS

Mice exhibited no significant difference among three groups in latencies to step through to the dark chamber of the passive avoidance equipment during learning in each session (table 1).

Prelearning treatment or postlearning treatment of mice with 10 times or 50 times MS did not impair passive avoidance response significantly, prelearning treatment or postlearning treatment of mice with ECS significantly impaired passive avoidance response in every session, as there was significantly reduction in day 2 test latency to step through (figure 1). There was no significant difference in latencies of test between prelearning treatment group and postlearning treatment group. And there was no significant difference in latencies of test between 10 times MS group and 50 times MS group.

DISCUSSION

In the present data from the passive avoidance method, the pulsed magnetic stimulation is considered to have no notable effect on the learning process of the passive avoidance, which ECS appears to effect the learning process. In our recent report [5)] of kinesiological neurochemical and neuropathological study from the effect of pulsed magnetic stimulation, there were observed that dopamine and serotonine metabolism were changed significanly after pulsed magnetic stimulation on rat's brain transiently, although there was no significant change in movement or neuropathology. The each pulsed magnetic stimulation excites the neurone, consequently the neurotransmitter is released from the synapse [5]. The observed changes of neurotransmitters are supposed to be due to the direct excitation of neurone by magnetic stimulation. During magnetic stimulation each muscle was contracted at every magnetic stimulation. However, after the cessation of stimulation, no special behavior or movement was observed at all [4]. These observation suggests that transient changes of neurotransmitter by magnetic stimulation does not interfere the memory, behavior or muscle contration, but further studies of long term momory, and of long exposure and strong exposure of magnetism should be necessary [11][12].

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

NORMAL RESPONSES AFTER MAGNETIC STIMULATION OF MOTOR CORTEX IN PATIENTS WITH SEVERE TETRASPASTICITY AFTER SUPRATENTORIC LESIONS

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INTRODUCTION

Good correlation of changes in evoked muscle responses after magnetoelectric stimulation of motor cortex with pathology changes have been shown in demyelinating disease (1, 2), in stroke (3, 4) and spinal cord lesions (5). From animal experiments (6, 7) and electrophysiological investigations it is highly probable that during the stimulation intracranial eddy currents stimulate the pyramidal neurons either indirectly via dendrites of interneurons or, especially when higher intensities are used, also directly at the PT-cell somata. Under this proposition, it is surprising to find almost normal MEP-results in patients with severe tetraspastic syndromes. Six cases are presented here: three after severe closed head injury, two after basilar artery occlusion and one after severe medicamental toxication, leading in all cases to a midbrain syndrome of various severity. All patients were investigated in an outpatient neurological rehabilitation clinic. Four of these six patients stayed for a rehabilitation treatment in the clinic, the remaining two were seen only once and did not participate in the rehabilitation program.

METHODS

Besides a thorough neurological examination the magnetic stimulation was done with a Cadwell MES 10 stimulator. The surface EMG was recorded from the thenar and abductor hallucis muscles bilaterally, using a 4-channel-EMG analysis system. For stimulation a 9 cm circular coil was positioned at the vertex, threshold was found starting at 40% and increasing in 10% steps of maximum intensity. All latency and amplitude measurements were measured with suprathreshold intensity either at 80 or 100% maximum intensity corresponding to 2.2 Tesla. The pulse shape was bipolar.

Degree of voluntary force was determined clinically and scaled after the MRC. Scoring zero for plegic and five for full strength.

RESULTS

The mean values averaged over all six patients were shown in Table I. The mean latencies of the patients are a little longer than normal but this difference ($\leq$ 2 sd), however, is not significant. One has to keep in mind, that it is nearly impossible to find a population of patients with diffuse brain lesion after prolonged coma without any additional focal lesions, which may be responsive for the little prolongation of mean latencies. In 23 out of 24 extremities a stimulation response is found despite the fact that in 14 limbs an MRC of 0 corresponding to complete plegia of the corresponding limb and in three more extremities a force score of one was present.

Table I: Group means of MEP-latencies and amplitudes versus voluntary force

	Force (MRC)	Mean latency (ms)	Mean Amplitude (µV)	Normals	
				Latency	Amplitude
left hand	1.83	22.9 ± 3.7	4216 ± 3500	20.7 ± 1.42	1770 ± 1200
right hand	1.50	23.4 ± 1.1	1685 ± 1350	21.0 ± 1.42	1570 ± 1000
left foot	0.50	48.7 ± 8.5	2366 ± 2250	42.4 ± 2.85	1260 ± 1300
right foot	0.80	48.2 ± 6.5	3116 ± 3840	42.5 ± 2.98	890 ± 800

DISCUSSION

From the clinical data it appears that the above results did not depend on a specific clinical diagnosis. A common clinical observation was, that all patients initially were comatose and developed midbrain symptoms of different degree. Duration of coma lasted from 3 to 6 weeks. In five cases CT-scans were made, showing as a common symptom signs of diffuse atrophy of white matter. Some few additional focal lesions were origin of additional side differences, but did not change the whole picture. In two patients an MRT-scan with high resolution and with high T2, weighting for better resolution of the midbrain, nuclei did not show any abnormalities in this region. There was especially no evidence of wallerian degeneration in the corticospinal tracts.

Assuming that diffuse axonal degeneration had occured in the patients of the above series (8, 9, 10), the following explanations of the results have to be considered:

1. Other than corticospinal neurons were stimulated. Stimulation of these neurons or axons may be facilitated by the enlarged cerebrospinal fluid volumes of high conductivity shunting eddy currents to deeper brainstem regions, which may show denervation hypersensibility in addition.

2. Some or all of the corticospinal neurons and their axons have survived or have regenerated, reestablishing a corticospinal connection but unable to reestablish normal motor function.

The possibility of stimulating other than pyramidal tracts cannot be excluded completely, because there are no exact theoretical models or experimental data describing the volumne conducted currents in normal and pathologic brains under magnetic stimulation. The pyramidal tract is the fastest known (60-80 m/sec) projection to the spinal cord it is known to be stimulated in normal subjects (11). So a stimulation of other descending tracts would be expected to result in longer latencies because of slower conduction velocities or because of additional synaptic delays.

In some studies (12, 4, 3) hemiplegic patients showed no or very delayed stimulation responses with diminished amplitude on the pathologic side, with no evidence of stimulation effects directly at brainstem regions, regardless of the size of the intracranial ischaemic defect.

So the most probable possibility is a remaining or reestablished corticospinal projection with near normal conduction velocity, in contrast to the apparent loss of voluntary motor function. So normal functioning of corticospinal neurons may be a necessary but not a sufficient condition for normal voluntary function.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

SELECTIVITY OF VOLUNTARY ENHANCEMENT OF MAGNETIC EVOKED MOTOR RESPONSES

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INTRODUCTION

Magnetic stimulation of motor cortex is of increasing importance as an clinical investigational method. A well known phenomenon here is a facilitatory effect on the motor evoked potentials (MEP) by voluntary activation of the muscle under investigation, resulting in a lower threshold, shorter latency and higher amplitude. In patients with hemiparesis, even in cases of absence of any voluntary EMG activity in the paretic muscles, a facilitatory effect by activating the unaffected side can be observed (1). So the question arose, whether this facilitation is due to a generalized central or peripheral activation effect or whether a selective facilitation of synergistic muscle groups is possible.

METHODS

The study was devided into two experiments with 10 participating subjects each. In the first experiment the subjects had the task to preactivate M. abductor pollicis, extensor digitorum, biceps and triceps brachii, respectively, as selectively as possible with about 10% of maximum isometric force of the respective muscles. The force was controlled by strain-gages, fixated with special splint, so that the motor actions were pure isometric. In both cases EMG was recorded by surface electrodes. The mean amplitudes of spontaneous EMG from bipolar derivations from muscles were derived from a continuous registration of EMG from the time interval 1 sec prior to the stimulation. The MEPs were recorded on a 4-channel-EMG-analyzer. Stimulus intensity was 5% above the highest threshold of the four muscles, but not more than 65% maximum stimulation intensity, which corresponds to 1.43 tesla. The second experiment had the same setup than the first one, except that the subjects were asked to activate the respective muscles not really but only mentally (i.e. just thinking about to activate them). The lack of overt contraction was controlled by the above mentioned strain-gages. The control condition in both experiments was complete relaxation of all muscles.

RESULTS

Experiment A: Preactivation

The latencies of all four muscles under the different conditions are shown in Table I. The latencies to activated muscles are selectively shorter. The amplitudes and the thresholds, as derived from the total number of responses, behave in a similar way. In order to quantify the selective activation, the selective condition was compared to the mean of the three other activation conditions, which showed latency decreases from 2.4 ms (extensor digitorum) up to 4.7 ms (thenar). All latency decrements are significant in t-tests ($p < 0.001$), as well as the amplitudes for the same conditions. To see whether there exists a global activation, the three non-selective activation conditions were compared to the relax-all-condition. Latencies as well as amplitudes did not differ significantly (unpaired t-test). Spontaneous EMG-activity, 1 sec prior to stimulation as a control, showed highly selective preactivation of the corresponding muscles. The amplitude differences were relatively higher than the MEP amplitudes.

TABLE I

Mean latencies and standard deviation in different muscles under selective preactivation

Latencies (ms)					
activate selectively	thenar	extensor	biceps	triceps	relax all muscles
thenar	**17.3 ± 4.0 n=96**	21.3 ± 3.3 n=70	22.4 ± 2.2 n=57	22.4 ± 1.6 n=80	22.5 ± 1.9 n=49
exten. dig.	17.6 ± 2.3 n=72	**15.1 ± 3.4 n=93**	17.6 ± 2.6 n=73	17.4 ± 1.6 n=85	17.3 ± 1.6 n=47
biceps	15.7 ± 1.8 n=17	16.0 ± 2.9 n=17	**10.9 ± 2.5 n=89**	14.5 ± 3.1 n=55	14.2 ± 1.3 n=26
triceps	16.1 ± 1.6 n=24	14.5 ± 3.7 n=17	16.1 ± 3.4 n=40	**10.9 ± 2.4 n=81**	16.7 ± 2.6 n=19

Experiment B: Mental activation

In this condition the recording of the spontaneous EMG prior to the stimulation shows no condition effect, except for the relax-all-condition, proving that the subjects really did not preactivate peripherally and (if at all) only mentally (see Table II). Nevertheless, the MEP amplitudes again are enhanced as shown in Table III and the latencies are shorter in the selective condition. These effects are much smaller than with muscle activation. T-test analysis shows significant differences ($p < 0.05$) only for biceps and triceps muscle.

TABLE II

Mean EMG and standard deviation in different muscles under mental activation

EMG (a.u.)					
activate selectively	thenar	extensor	biceps	triceps	relax all muscles
thenar	1.4 ± 0.8 n=55	**1.8 ± 2.0 n=57**	1.7 ± 1.6 n=56	1.7 ± 1.6 n=53	1.3 ± 0.4 n=53
exten. dig.	1.3 ± 1.0 n=55	**1.7 ± 1.2 n=57**	1.3 ± 0.5 n=56	1.3 ± 0.5 n=53	1.2 ± 0.5 n=53
biceps	2.1 ± 2.3 n=55	2.0 ± 2.0 n=57	**2.3 ± 2.6 n=56**	2.1 ± 2.1 n=53	1.4 ± 0.9 n=53
triceps	1.3 ± 0.5 n=55	**1.4 ± 0.5 n=57**	1.3 ± 0.5 n=56	1.3 ± 0.5 n=53	1.2 ± 0.4 n=53

TABLE III

Mean amplitudes and standard deviation in different muscles under mental activation

Amplitudes (µV)					
activate selectively	thenar	extensor	biceps	triceps	relax all muscles
thenar	**640 ± 463 n=55**	570 ± 565 n=57	480 ± 512 n=56	551 ± 500 n=53	670 ± 660 n=52
exten. dig.	520 ± 450 n=55	566 ± 458 n=57	**570 ± 490 n=56**	513 ± 360 n=51	480 ± 620 n=53
biceps	262 ± 494 n=54	250 ± 440 n=55	**420 ± 540 n=54**	180 ± 246 n=51	190 ± 230 n=48
triceps	126 ± 165 n=54	133 ± 125 n=53	138 ± 148 n=55	**200 ± 156 n=53**	120 ± 180 n=46

CONCLUSION

It is shown, that it is possible to facilitate different functions of muscle groups selectively by preactivation and to reduce the mean average latency up to 4.7 ms, which may well be sufficient to change the difference from pathologic to normal and vice versa. The facilitation effect may be explained by lowering the threshold of the α-motoneuron by additional activation through afferents. The results of premovement facilitation of MEP and Hoffmann's reflexes shortly before voluntary movements (2) also agree well with this hypothesis. The results of the second experiment show, however, that facilitatory effects cannot be explained alone by lowering the α-motoneuron threshold, because under this condition spontaneous prestimulus EMG activity should have changed also, which is not the case (Table II). Furthermore, MEP latencies and amplitudes did not correlate with the level of spontaneous EMG activity, neither with regard to the mean values nor with regard to the single stimulation responses. Possible mechanisms explaining this effect could be presynaptic effects either on spinal or on cortical level. Cortical facilitatory effects could be possible either by lowering the threshold of PT-neurons themselves or of afferent neurons, as by magnetic stimulation the PT-neurons are not stimulated directly but by intracortical synapses (3). The fact that facilitatory effects are also seen in direct stimulation of PT-neurons by electrical stimulation is not in contradiction to this possibility.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

EFFECTS OF THE TRANSCRANIAL MAGNETIC PULSE STIMULATION ON THE METAL USED FOR ORTHOPEDIC OR CARDIAC SURGERY

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INTRODUCTION

Recently, the transcranial masgnetic stimulation to the motor cortex has been widely used as a non-invasive technique for evaluating neural function, since the technique was reported by Barker in 1985 [1)-5)]. In Japan, Mano applied this technique for clinical use soon later on, and accumulated many data of normal and pathological conditions[6)-8)]. In this study, we examined the movements of iron particles by the magnetic stimulation for analyzing the magnetic fields in the case of smaller distance compared to the source size of magnetic flux. The safety aspects were discussed with respect to potential hazard in wearing metals or devices used for orthopedic or cardiac surgery through the transcranial application of pulse-shaped magnetic stimulus. An abstract of the preliminary data has been published[9)]

MATERIAL AND METHODS

Magnetic stimulator

Magnetic stimulator used in this study is of charge-discharge system with high voltage (800 V) and large capacity (1.5 mF). Maximum current I_{max} through coiled probe (its inductance 7μH) served as the source of magnetic flux was 8 kA with the rise-time of about 200 μs from the onset to I_{max}. The maximum strength of magnetic fields B_{max} was 2.34, 0.94, or 0.46 T according to the size of probe (52, 85, or 139 mm in mean diameter).

Magnetic Field

The magnetic field produced by the current passed through coiled probe was measured by the Gauss meter (Denshijiki Industry Co., Ltd., Tokyo, Japan; Model GM2430DS) with the the Hall-element (detector). The output of the Gauss meter was displayed on the conventional digital storage olloscope-screen, and plotted on X-Y chart with various distance regarding the geometry between the detector and the center of coil was changed in horizontal and vertical directions.

Iron particles

The macroscopic movements of the iron particles having the mean diameter 145μm, $FeCl_2$, $FeCl_3$, and stainless steel (SS) chips for general use were observed in varying strength of the magnetic stimulus and changing relative position between the detector and the stimulus coil. When the magnetic stimulus was applied on the iron particles scattered homogeneously over the acryl-plate, the moved iron particles were caught by dual adhesive tape attached to the coil plane.

Metals and Devices

The movement of the metals or the function of divices used for orthopedic or cardiac surgery, including bone screws, aneurysm clips, and multiprogramable pulse generators (cardiac pacemakers), was checked in vitro during the transcranial magnetic stimulation. Similarly, telephone cards, floppy discs, watches, and eye-glassess were also tested.

RESULTS

Magnetic Field

The relationship between the magnetic field stregth and the axial distance from the center of coil to the detector was shown in Fig.1. The difference between experimetal data and theoretical estimation with $B_x = \mu Ia^2 / 2(a^2 + x^2)^{3/2}$ was observed in this figure, where, B_x : the magnetic flux density at x of the deistance stated above, a : the diameter of the coil, μ : permeability.

Iron particles

There was no macroscopic movements observed over 300 mm from the probe-plane in any cases in this study. No iron particles were caught by the adhesive tape at probe-plane over 90 mm from the acryl-plate at B_{max} = 0.94 T. Typical movements of iron particles are shown in Fig.2. In addition, no macroscopic movements was observed in the any cases of $FeCl_2$ and $FeCl_3$ in this study.

Metals and Devices

Bone screw. No macroscopic motions were observed in the case of special stainless-steel (SS) [Sankin Kogyo Co., Ltd, Tokyo, Japan; SPM series], cobalt-chrome alloy (Co-Cr) [Robert Mathys Co., Bittlach, Switzerland; Synthes series, and Huwmedica GmbH, Schoenkirchen, BRD; Vitallium /Duo-driv series], and titanium (Ti) [Medicon GmbH, Tottoringen, BRD; Titanium- miniplatten system] at the magnetic stimulation of standard coil B_{max} .

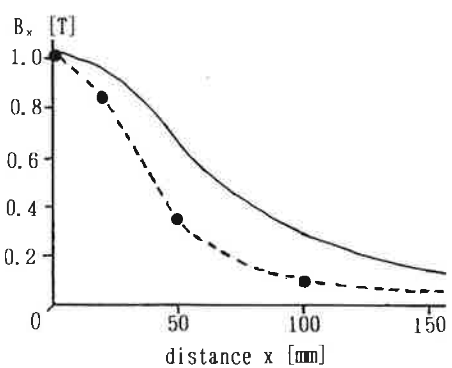

Fig.1 The magnetic flux density and the axial distance from the center of coil to the detector.
······· Experimental —— Theoretical

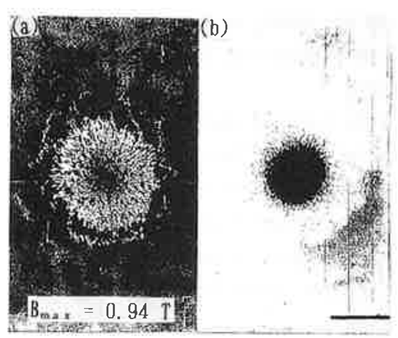

Fig.2 Movements of iron particles after the magnetic stimulation. Bar indicates the radius of the coil, 72.5 mm. (a) 50 mm from the center. (b) absorbed beneath the coil plane.

Aneurysm clip. No macroscopic motions of the Aneurysm clip [Mizuho Ikakogyo Co., Ltd., Tokyo, Japan; Sugita Standard Aneurysm Clip for Permanet Occulusion] were observed during the magnetic stimulation of standard coil B_{max} .

Cardiac pacemaker. Multiprogramable pulse generator (cardiac pacemaker), such as very popular one [Siemens-Elema AB, Solna, Sweden; DIALOG 728], was affected in its function with supressed output by exposing to the pulsed magnetic field of standard coil less than 1/10 B_{max}.

Others. Telephone cards became invalid, or not accessible by the card reader after exposure of pulse-shaped magnetic stimulus of standard coil 1/2 B_{max} . Flopy discs could be re-usable after correction of I/O error by reformating, whe* the discs were exposed to pulse-shaped magnetic stimulus of standard coil 1/4 B_{max} . Watches went wrong after the strongest magnetic flux with stardard ciol. Most types of eye-glassess were torelable.

DISCUSSION

The transcranial brain stimulation has become a very popular examination in neurology today, however there are some problems remaining unsolved; safety and unknown eddy currents inside the motor cortex. For safety aspects, Mano[10)-12)] and Matsumiya[13)] have recently reported the basic data including the histological and histochemical changes after multiple (over 50 times) transcranial exposure in animals at the supra-maximum strength (about 2.5 T) of that in clinical use, suggesting that the routine examination by transcranial magnetic pulse stimulation was tolerable.

It is recommended that this magnetic brain stimulation should be avoided in patients with epilepsy, severe psychiatric disease and wearing metals or electromagnetic devices[14)] . Mano[15)] has revealed the enhancement of epleptic discharges in experimental animals by the magnetic pulse brain stimulation.

Although there is no ferromagnetic substances inside the body which are macroscopically activated by the magnetic stimulation, there are many metals and electronic devices implanted into the body in recent years. In occasions, some of these metals are forced to move and/or functions of devices are affected, then it is necessary to have a special attention at the magnetic stimulation in patients with metals or electromagnetic devices.

Many types of bone screws have been developed and widely used in orthopedics or neurosurgery. These screws were made by special stainless-steel (SS), cobalt-chrome alloy (Co-Cr), or titanium (Ti). Aneurysm clip for permanent occlution is also widely used in cardio-vascular or neuro-surgery. These metals in the case of ferromagnetic substances may bring on troubles with halation or motion in the nuclear magnetic resonace system.

In recent years, multiprogrammable pulse generator has become the most common type of cardiac pacemakers for the treatment of arrythmia with A-V conduction block. This device can alternatively be set to demand mode or to fixed-rate mode by transthoracic application of the permanet magnet. It has been pointed out that the electromagnetic interference against the intra-ECG amplifier part of the pacemaker in demand mode changed to fixed-rate mode, causing fibrillation from R on T[17)]. Extremely large magnetic flux linkage with pacemaker leads, especially in unipolar system, might induce enough strength of electromotive force to stimulate the heart [18)].

There are experimental difficulties in observing the electromagnetic phenomena inside the body, leading discussions and studies on the identification of the stimulated site or on focusing techniques of the magnetic flux [19)-22)]. The optinum design and caculation of the magnetic flux are still problems[23) 24)].

ACKNOWLEDGEMENTS

We are grateful to Director, Masaru Yarita, Nihon Kohden Tomioka, Corporation, Tomioka, Japan, for his useful technical advices, and to Director Yoshiro Sawada, Kobe College of Medical Sciences and Medical Foundation Jikeikai Shinsuma Hospital, Kobe, Japan, for his giving an opportunity to present the study.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

THE ANALYSIS OF BALLISTIC MOVEMENT IN PATIENTS WITH CEREBELLAR ATAXIA

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INTRODUCTION

The various peripheral afferent inputs in ramp movement are known to be important factors of motor control. However, ballistic movement is not affected by such afferent inputs, whereas it is suspected of being controlled by so-called pre-programmed motor control. The cerebellum plays an important role in the programming process of the central nervous system. In this study, the accuracy of ballistic flexion movements of the knee joint was analyzed in patients with cerebellar ataxia.

SUBJECTS

The subjects were 13 patients with cerebellar ataxia, including 10 with spinocerebellar degeneration, 1 with cerebellar infarction, 1 with cerebellar hemorrhage and 1 with mercury intoxication. The neurological findings, activities of daily living (ADL) and ataxia stage were evaluated. The ADL test was established by the Research Committee for Neuromuscular Rehabilitation of the Japanese Ministry of Health. The test consisted of 32 items graded 0 to 3. The total ADL score in a normal subject was 96. None of the subjects showed any deficits in position sense, mental function or other higher cortical functions.

METHODS

The analysis of ballistic flexion movements of the knee in a supine position was composed of three sections. In the first section, as the target of knee flexion, the contralateral knee joint was flexed passively and the tested knee joint voluntarily flexed ballistically. In the second section, after the tested knee joint was flexed passively, the subject was asked to memorize the angle and the joint was extended. The tested knee joint was then flexed voluntarily and ballistically to the memorized target. In the first and the second sections, the target angles of knee flexion were : mild flexion (approximately 10 degrees), moderate flexion (35 degrees) and marked

target	mild flexion	moderate flexion	marked flexion
contralateral knee flexion	12.0 ± 9.1	11.8 ± 7.2*	3.4 ± 4.4
		**	
ipsilateral knee flexion	13.6 ± 6.5#	8.0 ± 6.1*#	4.3 ± 4.5
		**	

* $p < 0.02$
** $P < 0.01$
\# $P < 0.05$

Table 1 : Error (degree) in the angle of knee joint flexion.

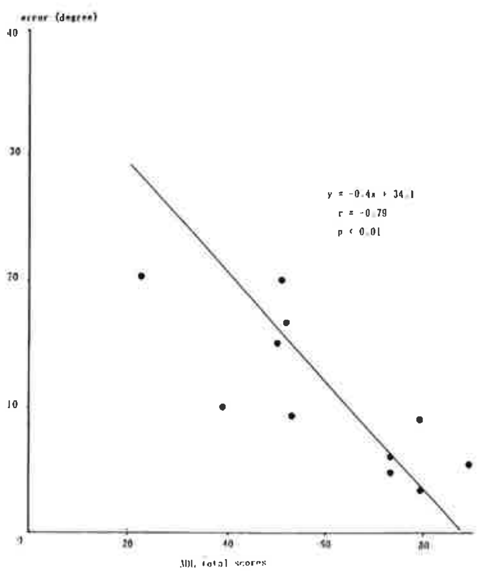

Figure 1 : Error (degree) in knee flexion and ADL scores.
target : contralateral flexion.

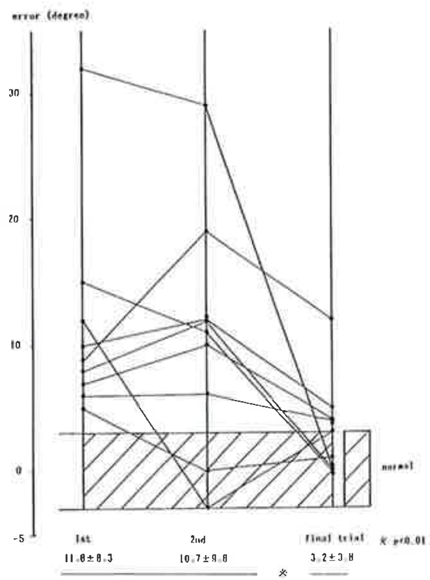

Figure 2 : Change in error (degree) with the repetition of moderate knee flexion.
target : contralateral flexion.

flexion (60 degrees). In the third section, ballistic voluntary knee flexion of a moderate angle was repeated. The error was measured as the difference between the target angle and the angle produced when tested. Reaction time and movement time were also measured in the first section. Each procedure was performed separately 1-2 months apart.

The apparatus used to record the movements included : a position system, a goniometer and a surface EMG with polygraphical recordings. The position sensor system (Hamamatsu Photonics Co.) has a highly sensitive detector (PSD) with a frequency of 300 Hz. The PSD is a planar type PIN silicon photodiode with very uniform layers formed on both the top and bottom surfaces. On the opposite edges of both surfaces, electrodes were provided to sense X and Y axis signals.

When the light spot from the light-emitting diodes , which are attached to the subject, are focused via a lens system on the active surface of the PSD, a photo current is generated at the spot position in the PSD. Portions of this current are collected at each of the electrodes through both resistive layers. The proportion of current at each electrode is determined by the resistance between the spot position and each electrode, i.e., the distance between them.

RESULTS

Hyperflexion was the error in all of the subjects. The error in marked flexion was smallest which was significant statistically at ($p < 0.01$) and the error in mild flexion was largest. The error in moderate flexion in the first section was significantly larger than in the second section ($p < 0.02$). There was a significant correlation between ADL scores and movement time ($y = -0.17x + 90.98$; $\gamma = -0.83$, $p < 0.01$). The errors were well correlated with ADL scores ($y = -0.4x + 34.1$; $\gamma = -0.79$, $p < 0.01$) and ataxia stage (Spearman rank correlation ; $\gamma = 0.89$, $p < 0.01$). However, there was no correlation between the error and reaction time or movement time. The severe ataxic cases were separated into two groups, with one group having a shorter movement time and more error and the other group having a longer movement time and more error. The latter group showed more disabilities in ADL and clinical findings than the former group.

Although all subjects showed error in the first trial, with the repetition of flexion all of the cases, except one who had a serious case of ataxia, improved their error by several repetitions of ballistic flexion.

DISCUSSION

When the target is contralateral knee flexion, the cerebellum controls to the motor program for planning the voluntary knee flexion movement. The other target was the memory in which the ipsilateral knee joint had been moved passively, which involves the motor programming. In the former test there was a target in the air, in the latter there was a target in the memory. The former was the passive movement of the contralateral knee and the latter was the passive movement of the ipsilateral knee.

In these different situations, the planning and the execution of movement was carried out. In ballistic small movement the information to the motor planning did not make a difference in the degree in error.

In ballistic moderate movement, the latter decreased the error more than the former. The change in the motor memory by the central nervous system may be suspected to affect the feedback system of the central nervous system in moderate ballistic knee flexion.

In marked flexion, errors in both groups decreased markedly. It is suspected that various peripheral feedback systems, like Ia and II fibers, may participate in motor control. These feedback mechanisms are considered to work on both groups.

With the repetition of movement, a decrease in error was observed. It is suspected that this can be contributed to motor learning. A severe lesion in the cerebellum may be suspected to impair motor learning. It is speculated that these mechanisms are related to neural plasticity which was observed as a long term depression in the cerebellum.

The degree of error was well correlated with ADL. The disability in ataxia was well correlated with error. The indicator of accuracy and error is thought to be an important factor in cerebellar lesions.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

FUNCTIONAL NEUROMUSCULAR STIMULATION THRESHOLD ELEVATION WITH FATIGUING PARALYZED MUSCLE

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INTRODUCTION

If functional neuromuscular stimulation (FNS) of paralyzed muscle is to become a useful technique for restoring function to persons with spinal cord injury (SCI), more needs to be known about the fatigue characteristics of these muscles and the corresponding changes in required FNS parameters. Since muscle contraction characteristics change markedly with continued activity (i.e., fatigue)[6], these data are important for the design of control algorithms that can adapt FNS parameters. Therefore, the purpose of this study was to document changes in threshold current, contraction force per unit of stimulation current, and maximal force output during repetitive FNS-induced contractions of paralyzed human quadriceps muscle.

METHODS

Testing was performed on the quadriceps muscle groups of 8 volunteer SCI subjects (age = 24-55 yr; C5-T9 with 4 complete and 4 incomplete lesions; time since injury = 1.5-16.8 yr) using a force-current measurement system[3] (Fig.1).

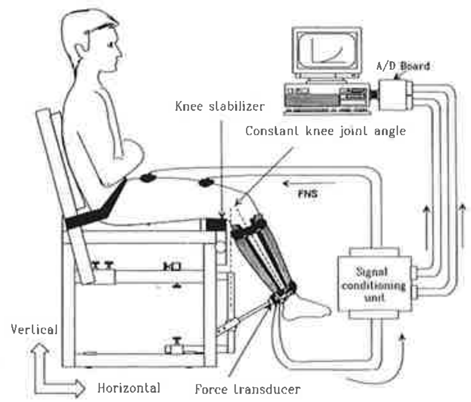

Figure 1. Schematic of the force-current system.

All participants gave informed consent to an institutionally approved protocol. Subjects were seated with the knee stabilized at 45 degrees of flexion. FNS current (300 microsec pulses, 35 Hz) was ramped up (10 mA/sec) through surface electrodes to a maximum force output load of 147 N, or to the 150 mA maximum current level (whichever occured first). FNS current was then ramped down (10 mA/sec) to 0 mA. Muscles were stimulated every 30 sec for 40 contractions, or until the muscles fatigued to 25% of original force output. For this study, contraction threshold (the initiation of force indicative of muscle contraction) was defined as 3.67N (2.5% of 147 N).

RESULTS AND DISCUSSION

Figure 2 plots the FNS current required to initate threshold contraction in relation to the number of contractions for each of the eight SCI subjects. For three subjects, threshold current increased markedly between contractions 1 and 6, and the threshold current increased gradually between contractions 11 and 36. All but one subject demonstrated a higher threshold current level at the end of 40 contractions when compared to the initial contraction. The bottom plot was from this one relatively strong subject whose knee extension force reached 147 N during all 40 contractions without marked fatigue and, therefore, no clear increase in threshold current was observed.

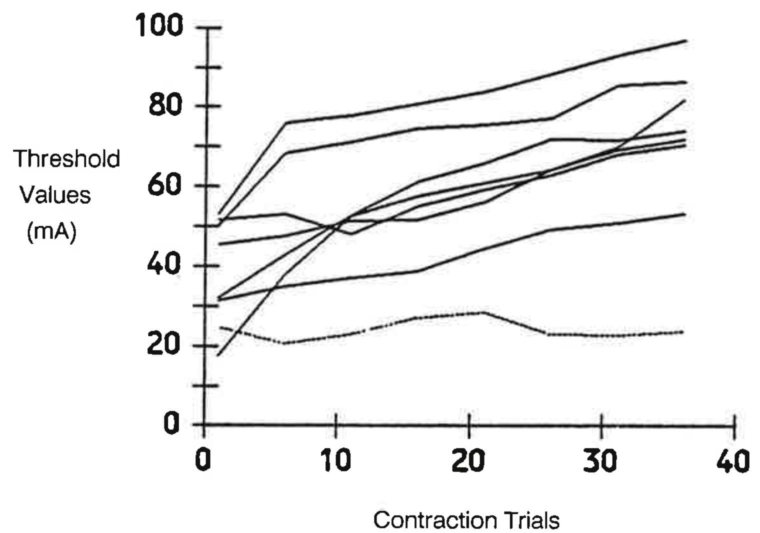

Figure 2. Current required to initiate threshold contraction in relation to the number of contractions.

Figure 3 illustrates the characteristics of a relatively weak quadriceps in one SCI subject. This muscle was able to achieve 147 N on only the first contraction, with lower maximal force output on subsequent contractions due to progressive fatigue of motor units. The threshold current ranged from 50 to 87 mA during contractions 1 through 36. This figure depicts three different fatigue indicators: 1) increasing threshold levels, 2) decreasing slope of the force per unit current, and 3) decreasing maximal force output.

Although the mechanism for fatigue is not entirely clear, it is feasible that the ramped FNS first excites the highly fatigable fast twitch (type II) motor units due to the lower threshold level of their large diameter motoneurons.[2] Higher levels of FNS current recruit the less fatigable slow twitch (type I) motor units whose motoneurons have higher thresholds due to their smaller diameter. Thus, as the initially stimulated motor units fatigue (and possibly drop out) with repetitive FNS, higher threshold current is required for contraction. As more motor units fatigue, less force per unit of current and less total

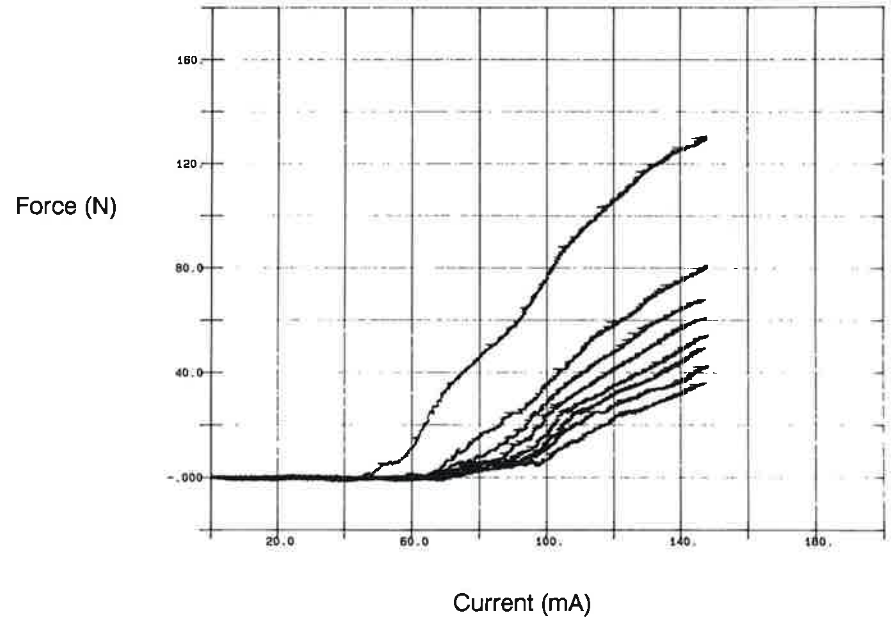

Figure 3. Force-current plots showing fatigue in one SCI subject. Every fifth contraction was plotted although a total of 40 contractions were accomplished by this subject.

force output are elicited. Therefore, these mechanisms of fatigue may include motor unit drop out and/or a weakening of the functioning motor units[4]. Further research investigating the relationship of fatigue measurements to time since injury, spasticity, lesion level, electrode placement[1], and current frequencies[5] is needed.

CONCLUSIONS

It appears that there is great variability in the contraction characteristics of paralyzed quadriceps muscles during FNS as illustrated by weak vs strong quadricep differences. These characteristics are influenced by the degree of fatigue as well as the basic fitness of the muscle. Mechanisms of fatigue may include motor unit drop out and/or a weakening of the functioning motor units.

ACKNOWLEDGEMENT

This study was supported by the Rehabilitation Research and Development Service of the U. S. Department of Veterans Affairs and Wright State University.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

THE INHIBITION OF MUSCLE ACTIVITY FROM THE CENTRAL NERVOUS SYSTEM IN RHEUMATOID ARTHRITIS WITH SPECIAL REFERENCE TO THE PATHOGENESIS OF MUSCLE ATROPHY

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INTRODUCTION

In patients with rheumatoid arthritis (RA), severe muscle atrophy and disturbance of motor function are often observed so that neurological disorders are suspected. However, the muscle atrophy is generally regarded as disuse atrophy resulting from motion inhibition due to pain and articular deformation. Disturbance of motor function in RA patients is also considered to be a secondary symptom similarly resulting from pain and articular deformation. These muscle atrophy and weakness seldom respond well to muscle strengthening exercise.

In an attempt to clarify the pathogenesis of muscle atrophy as found in RA patients this study was designed and compared with muscle atrophy of osteoarthritis (OA) patients.

SUBJECTS

The subjects were 12 female patients with RA and 14 female patients with OA. The mean age of RA was 53.1 years old and of OA was 50.4 years old. All RA patients were rated as Class 3, stage III according to Steinbrocker's classification. The subjects underwent total knee arthroplasty, immediately followed by repetitive articular passive motion using CPE (continuous passive motion exerciser). From the postoperative 2nd week, muscle strength evaluation and strengthening exercise was begun. The RA patients and OA patients received identical treatment.

METHOD

<u>Determination of EMG-reaction time and latency of knee stretch reflex</u>

In the present study, the latency from a stimulus signal to the onset of muscle discharge is called "premotor time", the time from a stimulus signal to initiation of leg motion is called "reaction time", and the time from muscle discharge to initiation of leg motion is called "motor time".[1)]

The electromyography aparatus employed was a DISA 1500 system to record EMG-reaction time. An acoustic signal transmitter were prepared. The subjects were made to extent their knee on hearing a buzzer. The electromyogram of the rectus femoris muscle was recorded by a pair of surface electrode. A light trigger generator was set to measure the time from transmission of an acoustic signal to the actual start of the leg motion which operated a light switch on and off. The latency of knee stretch reflex was measured by using a tendon hammer with trigger generator.

At each measuring session, the reaction time was determined 5 times after several trials by the subjects until they fully understood the procedure. These determinations were made only those patients who after more than 6 weeks after arthroplasty could easily bend and extent their knee.

Method of muscle strengthening and evaluation

Muscle strengthening and evaluation were carried out using an isokinetic muscle strengthening and evaluation device which was developed at our institution. Immediately after arthroplasty, the muscle strength was determined by the our device at the onset of the exercise, and the 1st, 2nd, 3rd, and 4th week after the start of the exercise.

We evaluated muscle strength as muscle recovery rate determined as follows:

Muscle recovery rate Ti/To - 1 (i=1,2,3,4)

To: muscle strength at the onset of exercise
Ti: muscle strength at the 1st, 2nd, 3rd, and 4th week after the onset of exercise

RESULTS

1. EMG-Reaction time and latency of knee stretch reflex

These results are summerized in the following tables (table 1,2,3 and 4).

TABLE 1. MOTOR TIME (msec)

	Operated Side	Control Side
RA Pt.	129.1 (SD 12.4)	129.7 (SD 28.6)
OA Pt.	124.7 (SD 32.5) *	113.5 (SD 28.6) *

* Difference between means assessed by Student's t-test, P<0.05.

TABLE 2. PREMOTOR TIME (msec)

	Operated Side	Control Side
RA Pt.	186.8 (SD 46.3)	165.0 (SD 47.1)
OA Pt.	165.2 (SD 49.2)	146.8 (SD 26.9)

TABLE 3. REACTION TIME

	Operated Side	Control Side
RA Pt.	314.9 (SD 38.4)	294.7 (SD 45.9) *
OA Pt.	289.9 (SD 69.2) *	260.4 (SD 49.0) *

* Difference between means assessed by Student's t-test, P<0.05.

TABLE 4. LATENCY OF KNEE STRETCH REFLEX (msec)

	Operated Side	Control Side
RA Pt.	21.2 (SD 1.2) *	20.7 (SD 1.5) *
OA Pt.	19.7 (SD 1.5) *	19.3 (SD 1.1) *

* Difference between means assessed by Student's t-test, P<0.05.

2. Muscle strength recovery rate

These results are below in the tables (table. 5 and 6).

TABLE 5. MUSCLE STRENGTH RECOVERY RATE IN RA PATIENTS

	1st week	2nd week	3rd week	4th week
Knee Extensor	0.18	0.21	0.32	0.55
Knee Flexor	-0.06	0.09	0.28	0.34

TABLE 6. MUSCLE STRENGTH RECOVERY RATE IN OA PATIENTS

	1st week	2nd week	3rd week	4th week
Knee Extensor	0.16	0.53	0.55	0.76
Knee Flexor	0.48	0.61	0.93	1.20

DISCUSSION AND CONCLUSION

On the basis of these findings, prolongation of EMG reaction time, and delay in the latency of knee stretch reflex, and low muscle strength recovery rate, it is considered that there is the inhibition of muscle activity from the central nervous system in RA patients. Moreover, in OA patients, a similar inhibition was speculated in the operated side, although to a mild degree.

According to histochemical studies on muscle in RA by Brook et al.[2], atrophy of type 2 fiber was observed in the early stage, whereas atrophy of type 1 fiber was found in the later stage. We believe these selective atrophy were caused by the CNS inhibition. Thus, in the early stage of RA, fine movements are impaired, and as the disease advances, the muscles necessary to maintain posture and position come to lose function.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

CHANGES IN MOTOR UNIT PROFILE IN PRIMATE EXPERIMENTAL HEMIPLEGIA (MORPHOMETRIC AND HISTOCHEMICAL STUDIES)

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INTRODUCTION

The motor unit profile of skeletal muscle is most susceptible to the pattern of its use (1). In the upper motor neuron lesions there is reduced motor unit recruitment and maximum firing rate (2). The physiological studies have further shown that in the flaccid state of hemiplegia the contraction time of the fast motor unit is increased without any change in twitch tension, while in the chronic state the fast units remain unchanged, but the slow units show increased twitch tension (3). Previous studies from our and other laboratories have demonstrated type 2 fiber atrophy along with type 1 fiber hypertrophy in cerebral and spinal spasticity (4,5,6). In spinal spasticity (7), studies failed to demonstrate any loss of spinal neurons arguing against transneuronal degeneration. Dietz et al (8) demonstrated normal reciprocal muscle activation in the spastic lower extremity, but the EMG amplitude was reduced. To our knowledge, this is the first study which looks chronologically at the motor unit profile of the long and short limb muscles following experimental hemiplegia.

METHODS

4 *Macaca Fascicularis* monkeys weighing 2.5 to 3.5kg were used for this study. Unilateral ablation was performed in the left cerebral hemisphere on cerebral cortical areas 4 and 6 of Brodman (9,10). The animals were sacrificed at 1 week, at 8 weeks and at 6 months after ablation. Two long muscles (biceps brachii and quadriceps) and 2 short muscles (thenar and lumbricals) were removed from both the left (control) and right (spastic) sides for histochemical and morphological studies.

ATPase staining at pH 10.2, 4.6 and 4.3 preincubation was carried out on 10-micron frozen sections for the typing of the fibers. Mean fiber diameter, distribution of the fiber types, fiber size ratio of the type 1 and 2 fibers were measured and the morphometric data from the control and spastic side were compared using the Bioquant Image Analysis system. Data were also collected and analysed on the fiber size distribution of the myelinated axons of the intramuscular nerves from the trichome-stained sections as well as from the 1-micron sections.

RESULTS

Clincal findings

The animals showed a dense right hemiplegia upon recovery from anaesthesia. During the third week after operation, the right limbs developed increased resistance to passive manipulation, consisting of mild plastic response to flexion or extension at the right shoulder, elbow, wrist, and knee. The deep tendon reflexes of the right limbs increased, so that they were brisker than those of the left limbs. In the fourth week, the right arm developed flexion movements at the shoulder and elbow in association with reaching movements of the lower limbs. The right limbs developed greater resistance to passive manipulation,and the deep tendon reflexes in these limbs became even more enhanced. During the eighth week, the right arm extended and the right hand made crude contact with a reflex hammer, coincident with the left hand, when the animal's chest was struck lightly with the hammer. At 6 months the animal showed spasticity. The increased resistance to passive manipulation and enhanced deep tendon reflexes persisted in the right limbs.

1 week after ablation

The long muscles (quadriceps and biceps) of the control side consisted of 23% type 1 fibers and 70% type 2A fibers and 7% type 2B fibers. The type 1 fibers ranged from 24-85 µm in diameter with a mean size of 45µm. The type 2 fibers ranged from 45 to 99µm with a mean of 68µm. The mean ratio of type 1 to type 2 fibers was 0.37.

The long muscles of the spastic side consisted of 20% type 1 fibers, 70% type 2A fibers and 10% type 2B fibers. The type 1 fibers showed a range of 22-70 µm in diameter with a mean of 45 µm. The type 2 fibers ranged from 25 - 100 µm with a mean of 60 µm. The mean fiber type ratio was 0.39.

The short muscles (thenar and lumbricals) of the control side consisted of 24% type 1 fibers, 70% type 2A fibers and 6% type 2B fibers. The type 1 fiber size ranged from 12-46µm with a mean of 23µm. The type 2 fibers ranged in size from 18-50µm with the mean diameter of 38µm. The mean fiber type ratio was 0.35. The short muscles of the spastic side consisted of 30% type 1, 60% of type 2A and 10% of type 2B fibers. The type 1 fibers ranged from 10-50µm in diameter with a mean of 24 µm. The type 2 fibers ranged in size from 18-68µm with mean of 30µm. The mean fiber ratio was 0.46.

In both long and short muscles in the 1 week animals the mean fiber type ratio of the spastic side was not significantly different from that of the control side. In the long and short muscles the type 1 fiber diameter was not significantly different between the control and spastic side. The type 2 fibers however were significantly smaller on the spastic side than the control in both muscles ($p<.001$)

8 weeks after ablation

The long muscles of the control side showed 20% type 1, 70% type 2A and 10% type 2B fibers. The type 1 fiber size ranged from 20-87µm in diameter with a mean of 44µm. The type 2 fibers ranged from 33-97µm with a mean of 63µm. The mean fiber type ratio was 0.23. On the spastic side, the muscles consisted of 30% type 1, 60% type 2A and 10% type 2B fibers. The diameter of the type 1 fibers ranged from18-59 with a mean of 40µm. The type 2 fibers showed a range of 23-86µm with a mean size of 56µm. The mean fiber type ratio was 0.43.

The short muscles of the control side had 20% type 1, 75% type 2A and 5% type 2B fibers. The mean fiber type ratio was 0.29. The type 1 fibers showed a size range of 12-40µm with a mean of 23µm. The type 2 fibers showed a range of 12-53µm with a mean size 0f 34µm. On the spastic side there were 24% type 1, 70% type 2A and 6% type 2B fibers. The type 1 fibers ranged from 11-40µm in diameter with a mean of 24µm in fiber size. The type 2 fibers ranged from 17-67µm with a mean of 35µm. The mean fiber type ratio was 0.25.

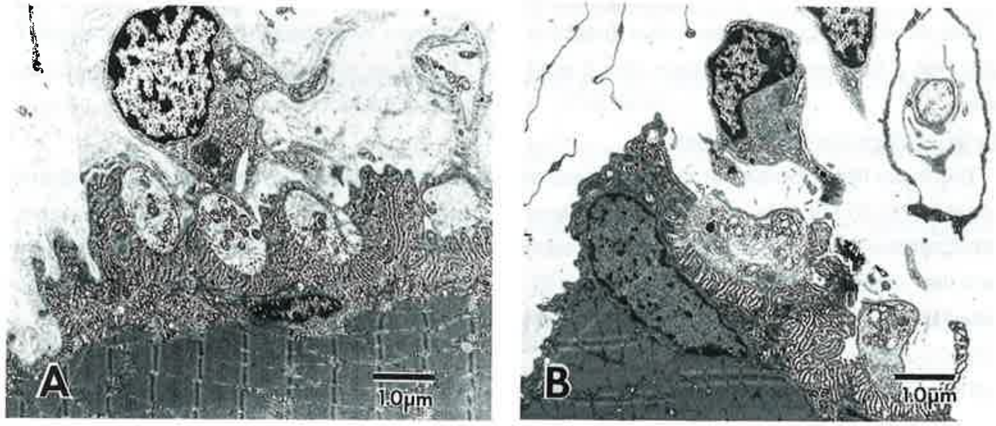

Fig.1 Morphology of motor endplates in the control (A) and 8 weeks hemiparetic (B) lumbrical muscle. Note the intact presynaptic and heavily folded postsynaptic regions.

The mean fiber type ratio of the spastic side did not show significant difference from the control side in both long and short muscles. In the long muscles, the fiber size of both type 1 and type 2 fibers was significantly ($p<.001$) smaller on the spastic side than the control side. In the short muscle, only the type 2 fibers showed significant size difference between the control and spastic side. ($p < .001$)

6 months after ablation

The control side long muscles showed 17% type 1 , 80% type 2A and 3% type 2B fibers. The mean fiber type ratio wae 0.20. The mean diameter of the fibers was type 1, 50µm (range17-89µm), type 2 ,59µm (range 17-110µm). The long muscles on the spastic side consisted of 27% type 1,67% type 2A and 6% type 2B fibers. The mean fiber type ratio was 0.44. The mean fiber size was type 1, 40µm(range 12 -62µm) type 2, 45 µm (range12-84 µm). The short muscles of the control side had 33% type 1 , 67% type 2A fibers. The mean fiber type ratio was 0.43. The mean diameter of the fibers was type 1, 26µm (range 14-45µm) type 2 36µm (range 18-81µm). On the spastic side the muscles consisted of 28% type 1, 65% type 2A and 7% type 2B fibers. The mean fiber type ratio was 0.59. The mean fiber size was type 1, 26µm (range 14-45µm) and type 2 36µm (range 21-64µm) in diameter.

Both type 1 and type 2 fibers of the long muscles of the spastic side showed a significant difference ($p< .001$) from that of the control side whereas in the short muscles only the type 2 fibers showed significant difference ($p<.001$) from that of the control. In both muscles the mean fiber type ratio did not show significant difference.

Comparison among spastic muscles

The type 1 and type 2 fibers of the long muscles of both the 8 weeks and 6 months spastic muscle were significantly smaller than the 1 week spastic muscle. The type 2 fibers of the 6 month spastic muscle were significantly smaller than the 8 weeks spastic muscle, where as the type 1 was not.

One week after ablation type 2 fiber atrophy was noted in the long and short muscles. Both type 1 and type 2 fiber atrophy was seen after 8 weeks and 6 months in the long muscles on the spastic side.

Nerve and endplate Morphometry

The mean fiber diameter of the myelinated axons of the intramuscular nerves of the control and spastic side did not show any significant difference. Morphometry of the endplate regions revealed, no significant difference in the mean presynaptic axon terminal area of the spastic muscle (6.07μm) and the control (4.68μm). The mean area of the postsynaptic folds also showed no difference between the spastic (6.76μm) and the control (4.76μm) sides (Fig. 1).

CONCLUSION

The present study confirmed Type 2 atrophy but failed to demonstrate any differences in the fiber type ratios of the muscles on the spastic and normal side, indicating preservation of the motor unit pattern in the spastic muscles over a 6 month period. The fiber size, however, showed reduction in the size of type 2 fibers at the 1 week, 8 weeks and 6 months in the long and short muscles. Type 1 fiber atrophy was seen at 8 weeks and 6 months of the long muscles.This may be accounted for by the disuse and may explain Meyer & Young's (3) observation of reduced amplitude of the type 2 units, and Tang and Rymer's (11) observation of increased ratio of EMG to isometric force on the paretic side.

The motor endplate size and intramuscular nerve distribution patterns were similar in both the spastic and control side in both the type 1 and type 2 fibers, again indicating that these subcomponents of the motor unit were unaffected.

From these findings we conclude that the motor units in experimental hemiplegia were unchanged up to 6 months. These findings have definite therapeutic implications. The observation further strengthens the pathological studies of Khoubesserian et al (12), and Kaelan et al (7), Oppenheimer (13) which failed to demonstrate any transynaptic degeneration of anterior horn cells after cortical spinal tract lesions.

ACKNOWLEDGEMENT

We are grateful to Dr. Gilman and Dr. Dauth (University of Michigan) for providing the muscle samples.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

ISOKINETIC EVALUATION OF MUSCLE POWER IN NEUROLOGICAL DISEASES

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INTRODUCTION

Functional assessment of patients affected by neurological diseases is difficult to objectify in the clinical practice, on the basis of neurological signs and observation of patient's global activities (e.g. deambulation). Objective evidence of diseases evolution and efficacy of pharmacological and/or training programs become thus difficult. Therefore an instrumental quantitative functional assessment becomes necessary.

Isokinetic tests have found wide application for evaluating muscular performance as regards sports medicine and sequelae of injuries of osteomuscular apparatus. We have verified the utility of isokinetic test of quadriceps and hamstrings during follow-up in patients affected by deambulation desorders related to SCI, S.Guillain Barré, Parkinson's disease; we also have a tool for provide objective evidence about pharmacological treatments or training programs efficacy.

MATERIAL AND METHODS

All patients have been submitted to isokinetic evaluation of right and left quadriceps and hamstrings at 60°-120°-180°/second angular speed using a Cybex II dynamometer. Peak torque; Time Rate of Tension Development (TRTD) to peak torque at 60°/sec; Reciprocal Inervation Time (RIT) at 60°/sec were evaluated.

SCI: 7 patients, aged from 21 to 60 years (mean 43 years): 3 evaluated before and after treatment with antispastic drugs (baclofen or tizanidine); 2 monitored during follow-up; 2 evaluated before and after isokinetic training of extensors/flexors knee muscles.

S.Guillain Barré: 3 patient aged from 52 to 64 years (mean 59 years): 1 monitored during follow-up; 2 evaluated before and after isokinetic training of extensors/flexors knee muscles.

Parkinson's disease: 6 patients aged from 53 to 68 years (mean 61 years) evaluated before and after pharmacological treatment.

TABLE I

PEAK TORQUE % INCREASE IN GUILLAIN BARRE' PATIENTS

Patient		60°/sec R	60°/sec L	120°/sec R	120°/sec L	180°/sec R	180°/sec L
1	EXT	15	22	25	38	55	40
	FLEX	0	11	4	20	14	13
2*	EXT	77	53	54	41	55	33
	FLEX	41	66	66	60	44	55
3*	EXT	937	257	400	157	227	275
	FLEX	231	200	400	327	223	616

R= right side L= left side *= training

TABLE 2

PEAK TORQUE % INCREASE IN PARKINSON'S PATIENTS

Patient		60°/sec R	60°/sec L	120°/sec R	120°/sec L	180°/sec R	180°/sec L
1	EXT	28	50	33	60	62	150
	FLEX	18	30	50	28	0	57
2	EXT	22	37	0	4	50	16
	FLEX	22	10	25	26	14	25
3	EXT	15	4	12	0	36	7
	FLEX	30	15	0	20	0	0
4	EXT	26	13	70	33	85	83
	FLEX	10	22	60	28	100	40
5	EXT	118	328	340	300	480	450
	FLEX	108	242	200	200	300	180
6	EXT	16	23	-10	-14	-7	-12
	FLEX	0	83	-7	66	0	14

FIG.1

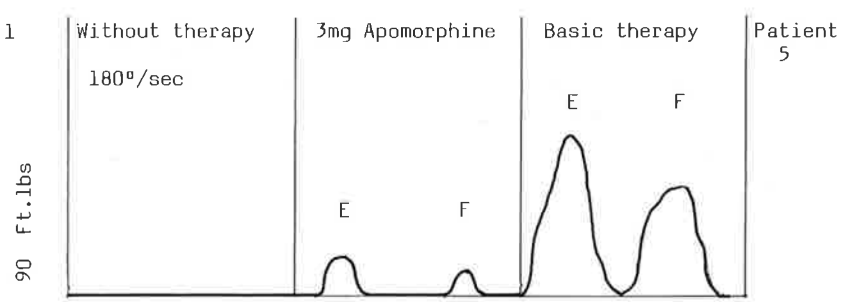


TABLE 3

PEAK TORQUE % INCREASE IN SCI PATIENTS

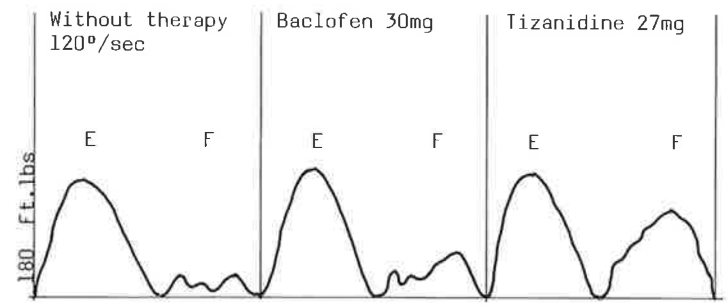
Patient		60°/sec R	60°/sec L	120°/sec R	120°/sec L	180°/sec R	180°/sec L
1	EXT	0	5	11	0	0	0
	FLEX	50	9	80	30	50	60
2	EXT	11	91	28	25	44	38
	FLEX	20	66	57	57	94	108
3	EXT	212	310	612	233	0	0
	FLEX	600	50	0	0	0	0
4	EXT	73	92	176	150	400	142
	FLEX	250	46	360	77	210	114
5	EXT	-20	33	-30	-5	-25	0
	FLEX	-8	76	-15	0	-26	18
6	EXT	-12	4	-46	0	0	0
	FLEX	0	-6	-18	0	0	28
7	EXT	0	0	26	6	33	0
	FLEX	0	-12	6	0	20	0

R= right side L= left side

Patient 1-2-3: pharmacological treatment (baclofen/tizanidine)

Patient 4-7 : training Patient 5-6 : follow-up

FIG.2: Patient 1



RESULTS AND DISCUSSION

Quadriceps and hamstrings isokinetic evaluation revealed realizable and repeatable in these neurological patients.The utility of isokinetic muscle parameters in the clinical practice seems to be noticeably interesting because with a relatively quick method you can obtain objective parameters for follow-up and therapeutic decisions.In fact in patients affected by secondary paraparesis related to SCI you can confirm antispastic drugs efficacy in spasticity control.We also monitored the effects of isokinetic training of E/F knee muscles in patients affected by SCI,S.Guillain Barré.
In Parkinson's patients it has been possible to give a more specific and effective pharmacological treatment by demonstrating qualitatively and quantitatively responses to pharmacological treatment.There is also the possibility to optimize the dosage of pharmacological treatment.It is reasonable a routinary use of isokinetic evaluation of E/F knee muscles in patients affected by deambulation disorders (perhaps not only in SCI,S.Guillain Barré,Parkinson's disease) as extension of clinical evaluation to obtain objective evidence and guidelines for functional assessment and treatment.

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Electromyographical Kinesiology
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ANALYSIS OF SHORT TERM MOTOR LEARNING IN IMPAIRED COORDINATION

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INTRODUCTION

It is often experienced that motor skill is clinically improved by repetitive performance of the movement in patients with impaired coordination as spinocerebellar degeneration. The role of the cerebellar system in motor learning has been studied (1-4), but quantitative evaluation on motor learning in impaired coordination is not so many (2). In this report, quantitative evaluation on impaired coordination of upper limbs was attempted, using an untouched type position sensor system, and motor learning in the short term within a few minutes was studied in patients with spinocerebellar degeneration.

SUBJECTS AND METHODS

Subjects were eight patients with spinocerebellar degeneration who consisted of three patients with late cortical cerebellar atrophy (LCCA), two patients with olivo-ponto-cerebellar atrophy (OPCA) and three patients with Holmes type hereditary cerebellar ataxia (Holmes), aged 42-71 (mean 55.9) years. These patients were diagnosed by clinical signs and radiological findings, and had no obvious intelligence disorder. Five age-matched normal controls, who had no neurological signs and consisted of four men and one woman, aged 41-62 (mean 51.8) years, participated in this study.

An untouched type position sensor system was used in this study. It consisted of a small object with a light-emitting diode (LED) and a position sensor camera. The LED could be moved freely. It gave off the light to the position sensor camera, and the sensor camera measured the position of the object with the LED two-dimensionally. Measured data by the position sensor were sent to a personal computer system and the two-dimensional trajectory of the object with the LED was expressed on the computer display by the programming. The length, the velocity, etc. of the trajectory was able to be indicated by the computer.

A subject sat in front of the plate glass which was inclined toward the subject from the horizontal. The subject moved an object similar to a cap (Cap in the following) by the hand from a home position on the plate glass to a goal position on the same one. The distance of the two positions was 250 mm and the home position was near the subject and the goal position was away from the subject. Movement of the Cap was begun by the trigger of the buzzer. The

Table 1 Mean ±SD of $L_{\%}$, D, V_{max}, MT in all trials

		$L_{\%}$ (%)	D (mm)	V_{max} (mm/s)	MT (msec)
Case 1	R	115.7± 7.0	11.4± 5.1	486±136	1360±391
	L	107.8± 3.6	41.0±10.6	607± 84	876± 73
2	R	110.7± 5.2	27.9± 2.6	1383±127	538± 37
	L	107.2± 1.2	24.9± 4.2	1456±102	561± 17
3	R	112.2± 3.6	13.5± 5.0	886±112	721± 68
	L	110.9± 7.2	25.5± 6.9	850±114	691± 77
4	R	130.6±19.1	21.1±11.1	822±223	1149±184
	L	127.1± 8.5	25.7± 5.8	633±253	1165±162
5	R	103.5± 1.7*	4.6± 4.8*	1220± 74	493± 29
	L	106.0± 1.8*	5.1± 3.8*	1563±144*	423± 63*
6	R	110.9± 5.2	11.4± 6.4	1038± 77	705±114
	L	111.1± 4.3	17.0± 9.0	969±169	657±121
7	R	109.8± 2.4	5.5± 3.4*	963±110	770± 97
	L	109.4± 1.8	5.6± 2.5*	1028±123	742± 49
8	R	109.9± 6.6	15.3± 2.1	1126±244	879±139
	L	112.7± 3.9	18.1± 2.6	966±148	840±130
Controls	R, L	104.8± 2.2	6.2± 4.0	1603±206	428± 44

* not significant compared with the normal group

subject lifted the Cap from the home position on the plate glass by one hand and transferred the Cap toward the goal position. The subject kept the Cap above the plate glass during the transfer. Finally the subject took down the Cap to the goal position. Movement of the Cap from the home position to the goal position was defined as one trial. The subject moved the Cap at the maximal velocity at first and moreover tried to do accurately. The subject did not practice before trials, but had the Cap and recognized the light weight of the Cap. LED was installed at the bottom of the Cap and gave off the light through the plate glass. The trajectory of the LED (i.e. Cap), which corresponded to the projection onto the plate glass, was shown on the display.

Trials were performed 10 to 15 times by the right hand at first and then 10 to 15 times by the left hand. The interval of each trial was about 15 seconds and the recording time of one trial was 1 to 3 seconds.

We studied about the length and form of the trajectory of the Cap projected onto the plate glass, the movement time (MT), the maximal velocity (V_{max}), the difference between the goal position and the reached position of the Cap (D) and the percentage of the length of the trajectory to that of the straight line from the home position to the reached position of the Cap ($L_{\%}$).

RESULTS

Means of $L_{\%}$, D, V_{max} and MT in all trials of each subject were shown in **Table 1.** Almost patients had increased $L_{\%}$ and D, prolonged MT and decreased V_{max} compared with the controls. But in case 5 and case 7 which had clinically very mild or mild ataxia of upper limbs, some of the means of $L_{\%}$, D, V_{max} and MT did not have significant differences. D, V_{max} and MT did not change significantly with repetitive trials in each subject.

In controls $L_{\%}$ was low from the first trial and always under 110 percents through all trials. In 5 patients (9 sides) $L_{\%}$ was high and variable in the early stage of repetitive trials, but $L_{\%}$ was decreasing and invariable in the end stage (**Figure**). In 2 patients (4 sides), who had clinically moderate to severe ataxia, $L_{\%}$ did not decrease with repetitive trials. 2 patients (3 sides), who had very mild ataxia, had the same course of $L_{\%}$ as the controls.

DISCUSSION

It is known that speed and accuracy are incompatible in the movement (3). In our study, we need speed to the subjects at first and accuracy secondarily. As the limitation of the movement was little because of an untouched type position sensor system, we were able to record the movement in more natural condition.

Figure Changes of $L_{\%}$ in Case 6 and Case 7

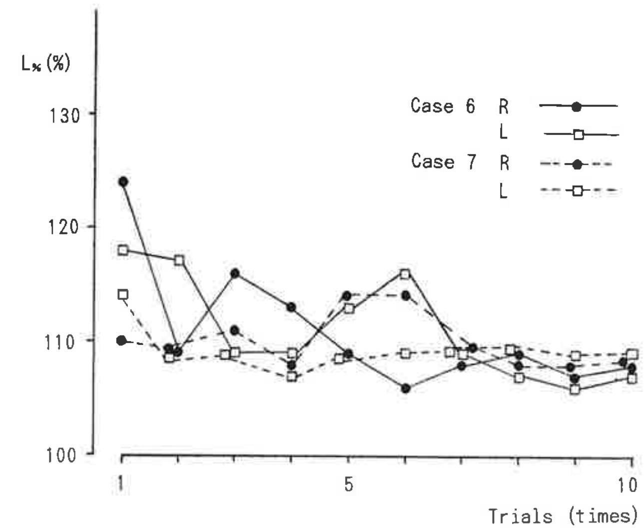

Table 2 Effect of short term motor learning concluded from decrease of L% with repetitive trials

Case	Age	Sex	Course	Diagnosis		Short term motor learning	Ataxia
1	50	M	9Y	LCCA	R	○	moderate
					L	○	mild
2	53	M	6Y	LCCA	R	○	moderate
					L	—	very mild
3	71	F	25Y	LCCA	R	×	moderate
					L	×	moderate
4	56	M	4Y	OPCA	R	×	moderate~severe
					L	×	moderate~severe
5	62	M	2Y	OPCA	R	—	very mild
					L	—	very mild
6	65	F	15Y	Holmes	R	○	moderate
					L	○	moderate
7	42	M	3Y	Holmes	R	○	mild
					L	○	very mild
8	48	M	30Y	Holmes	R	○	mild
					L	○	mild

○ recognized × not recognized — same as controls

It is considered that L% represents clinical ataxia of the upper limbs, especially decomposition. Decreasing and invariable L% with repetitive trials is regarded as the effect of short term motor learning (**Table 2**). When the subject moves the Cap, the program of the movement is decided on the basis of vision, and he moves the Cap by feedforward and feedback control. The program is corrected on the basis of the result of the trial and the next trial is performed. The plasticity of the motor control system as the above results in the effect of short term motor learning (4). As the effect of motor learning was not recognized in patients with severe ataxia in our study, it is considered that short term motor learning requires at least the remaining of the cerebellar function to some degree.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

LONG-TERM AMBULATORY RECORDING OF SURFACE EMG IN THE ASSESSMENT OF SPASTICITY

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INTRODUCTION

There is a clinical demand for tools for objective and quantitative assessment of spasticity. Various measurement systems have been proposed in literature [1, 2, 3, 4], but the great majority of these measure only variables describing the momentary state of spasticity. The clinician is not likely to accept such systems because spasticity is known to exhibit great time variability. Besides, most proposed systems were designed for use in the laboratory: the clinician questions the validity of the results of such laboratory tests for his clinical diagnosis and decision making, because spasticity is strongly affected by environmental factors.

In an attempt to avoid these problems, the use of long-term ambulatory recording of surface EMG in the assessment of spasticity was proposed [5]. This paper describes some properties of the measurement system which is being developed presently.

INSTRUMENTATION

The technique involves recording of surface EMG of two muscles using a computer-based, portable datalogger which is carried by the patient during his activities of daily living. For a block diagram of the portable system, refer to Fig. 1. The EMG is amplified and subsequently high-pass filtered for suppression of movement and cable artifacts (third order Butterworth filter, -3 dB at 20 Hz) . The filter's output signal is rectified and integrated over a user-defined time interval. The integrator output level is AD converted at the end of this interval and stored in the data-logger's memory, after which the integrator is reset. The recording lasts 24 hours. After completing the recording period, the stored data are off-loaded to a personal computer for data analysis.

Figure 2 presents a typical recording obtained from a patient with a complete spinal cord injury.

The patient keeps a diary of his activities of daily living to facilitate interpretation of the EMG recording.

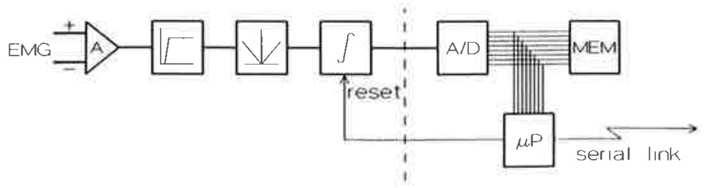

Fig. 1. Block diagram of portable ambulatory EMG surface EMG recording unit.

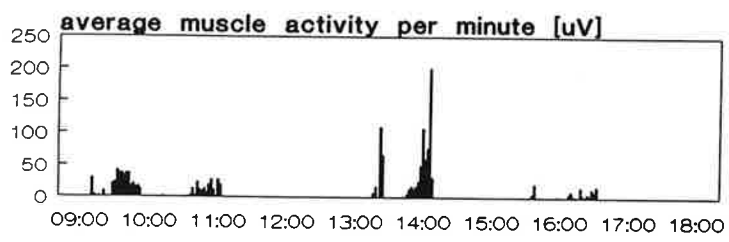

Fig. 2. Typical recording, obtained from complete spinal cord injured patient. Average muscle activity [microvolts] versus time of day [hours:minutes].

LONG-TERM STABILITY OF THE ELECTRODE-SKIN CONTACT

Crucial for the reliability of long-term EMG recording is the stability of the electrode-skin contact. Deterioration of the quality of this contact may result in a varying sensitivity of the measurement system. Therefore, this stability was tested for Medi-Trace pellet electrodes (Graphic Controls Canada Ltd.).

To test this stability, an EMG source which is highly stable over 24 hours is required. During measurements with the gait analysis system in our rehabilitation clinic [6], computer averaged activation profiles (rectified and smoothed EMG, synchronised with foot contact) of leg muscles are calculated from recordings of muscle activity of several gait cycles (the period between two heel strikes of one foot). It was found from these measurements, that the average activation pattern of the M. Rectus femoris of normal subjects during gait at comfortable speed is highly reproducible. For the experiments described below, the mean value of the average activation pattern of the M. Rectus femoris was used as a parameter. This parameter is identical to the integrated muscle activity as recorded during long-term measurements.

Four normal subjects walked a distance of 40 meters (corresponding to 12 to 20 recorded gait cycles) at comfortable speed in our gait analysis lab for 20 times. The standard deviation of the 20 parameter values of the recordings was less than 5 per cent of the mean. Four times within the 24 hours of the experiment (0, 4, 8 and 24 hours after the surface electrodes were placed on the skin) the measurements were repeated. In this way, 4 sets of 20 data were obtained for each subject.

In figure 3, the course of the mean value of the calculated parameter during the 24 hours of the experiment is displayed for each of the four subjects. No overall change in the recorded EMG activity is observed, which indicates that no significant change in the quality of the electrode skin contact occurs within the 24 hours of a long-term recording.

CONCLUSION

These results indicate that reliable long-term EMG recording is technically feasable. Because muscle activity is only one of many aspects of spasticity, further investigation of the clinical value of this technique is necessary. However, preliminary results of such recordings in complete spinal cord injured patients show that spasms, as noted by the patient in the diary coincide with the occurrence of EMG activity. This justifies further development of long-term ambulatory EMG recording for assessment of spasticity.

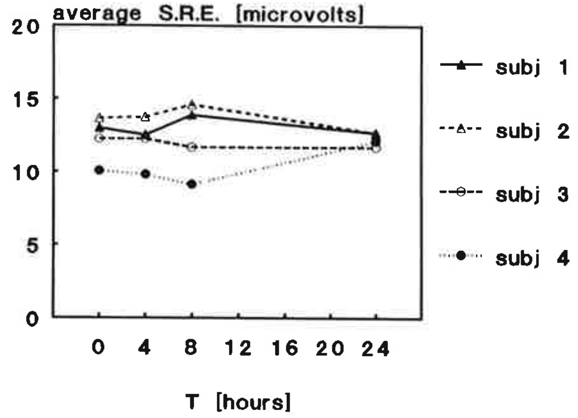

Fig. 3. Course of the average muscle activity per gait cycle of the M. Rectus femoris during 24 hours. Recordings were made at comfortable walking speed from 4 normal subjects. Symbols indicate the mean value of the average muscle activity.

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Electromyographical Kinesiology
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CHARACTERISTICS OF FUNCTIONAL NEUROMUSCULAR STIMULATION-INDUCED SPASTICITY IN SPINAL CORD INJURED SUBJECTS

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INTRODUCTION

An increasing number of researchers are developing functional neuromuscular stimulation (FNS) systems for the control of paralyzed muscle. However, the spasticity often associated with spinal cord injury (SCI) presents unique problems for FNS control systems. Spasticity is typically defined as a motor disorder characterized by a velocity-dependent increase in tonic stretch reflexes.[1] Although normal muscle response to electrical stimulation is well documented, the response characteristics of denervated and/or spastic muscle have not been adequately explored. The purpose of this study was to document the response to FNS of paralyzed, spastic muscle in individuals with SCI.

METHODS

Participants in this study were 19 volunteer SCI individuals who gave informed consent to an institutionally approved protocol. This heterogeneous group of subjects included 2 females, 17 males, 12 quadriplegic, and 7 paraplegics with lesion levels ranging from C_5 to T_{11}. Mean subject age was 31 $\pm$ 8 years, and time since injury was 7 $\pm$ 5 years.

Fig. 1. Force current measurement system including force transducer, surface electrodes, stimulator, computer for collection and analysis, and x-y plotter.

Testing for evidence of spasticity was performed on the quadriceps muscle groups of the subjects using a force-current measurement system (Fig. 1).[2] Subjects were seated with the knee stabilized at 45 degrees of flexion. FNS current (300 microsec pulses, 35 Hz) was ramped up through surface electrodes to a maximum load of 147 N, or to the 150 mA maximum current level, and then ramped down every 30 seconds for 40 isometric contractions or until the muscle fatigued to 25% of its original force output.

A computer graph showing eight plots of current level and resulting force output was obtained for each testing session. If 40 repetitions were completed, every fifth contraction was plotted, for a total of eight plots. Graphs were examined for rapid and abrupt force increases as evidence of muscle spasm.

RESULTS AND DISCUSSION

Out of the initial group of 19, 9 SCI subjects (or 47% of the total) were found to demonstrate spasticity with FNS. This sub-group (SP) included 5 quadriplegics and 4 paraplegics (lesion range C_5 to T_9), age 29 $\pm$ 5 years, with a mean time since injury of 7 $\pm$ 5 years. Without exception, those quadriceps muscle groups with spasticity generated higher isometric force for more repetitions than those without spasticity. Four of the 9 subjects with spasticity (44%) completed all 40 FNS contractions at the maximum load level. This seems to indicate a greater capacity for force generation in the SP group, regardless of spinal cord lesion level.

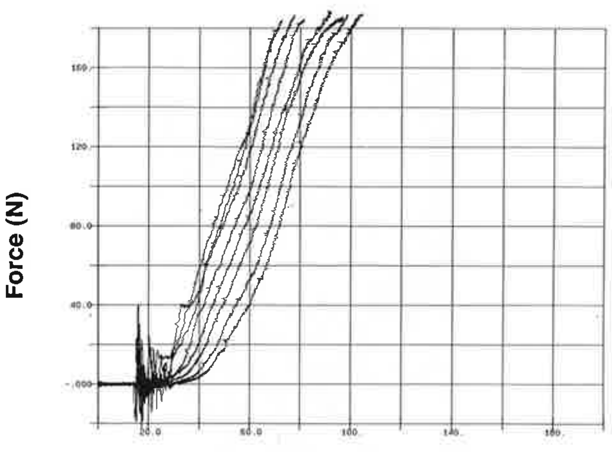

Fig. 2. Force-current plots showing the quadriceps spasticity response from one SCI subject initiated at low stimulation levels (low threshold excitation). Spasms were present throughout the 40 trials.

In five of the 9 SP subjects (55%), the muscle spasm was initiated at very low stimulation levels (<60 mA). This spasticity response (shown in Fig. 2) occurred at a lower level of stimulation than usual for FNS-elicited muscle contraction. A variety of possible neural mechanisms for spasticity have been proposed in the literature.[3] This type of response to *low* levels of stimulation seems to indicate an increased responsiveness of interposed excitatory interneurons to muscle afferent input.[4] This increased responsiveness may be caused by collateral sprouting of the terminal endings of muscle afferents to accommodate synaptic sites vacated by destruction of supraspinal tracts following SCI.[5]

Spasm was initiated at approximately 100 mA for one subject and at near maximum stimulation (150 mA) for two subjects. These stimulation levels are within the range usually found to elicit muscle contraction. This type of increased motoneuronal response to stimulation *after* muscle contraction has been initiated may be secondary to augmented stretch-evoked excitatory synaptic input. Some of the muscle spindle afferents may show enhanced response to stretch which occurs *within* the isometrically contracting muscle.[6]

Fig. 3 shows the spasm onset for the remaining subject (C6 incomplete quadriplegic). Low FNS/current levels and subnormal contraction threshold triggered spasms initially, but these subsided with repeated contractions. However, a second spasm which was triggered at high FNS/force levels was present for all 40 repetitions. This subject may demonstrate both increased responsiveness to stimulation via a lower threshold required for excitation and enhanced stretch-evoked synaptic excitation of motoneurons.

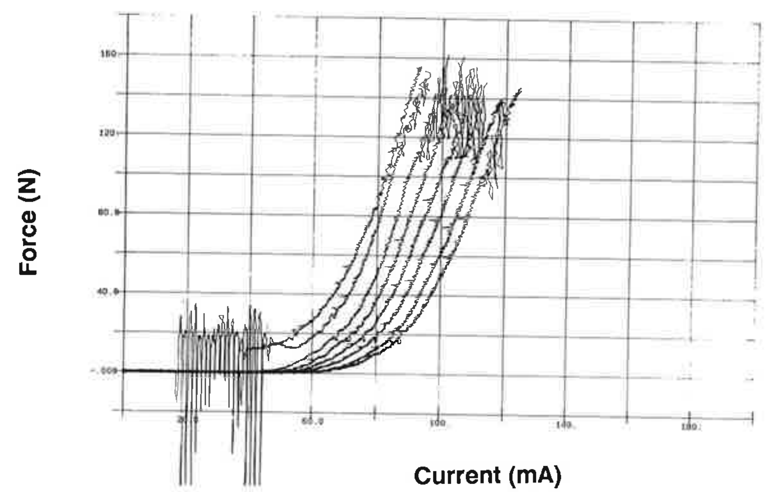

Fig. 3. Force-current plots from the quadriceps muscle group of one subject who demonstrated a combination of both low and high threshold spasticity responses to stimulation. Although the high threshold spasticity response was present throughout the 40 trials, the low threshold response was present only in the first 10 trials.

In 7 of the 9 SP subjects (78%), the spasms decreased or completely stopped with repeated stimulation. This may indicate that the excitatory interneurons which are more sensitive to muscle afferent stimulation or to stretch-evoked excitation actually fatigue with repeated stimulation. Repeated measurements on 2 SP subjects demonstrated variability in stimulation level for spasm initiation. Although this may indicate that the threshold for motoneuronal excitability or stretch-evoked excitation varies, it may also reflect variability of electrode placement, temperature, medication, skin impedance, or state of muscle fatigue.

CONCLUSIONS

Results using a force-current measurement system show a greater capacity for isometric muscle force generation in paralyzed muscle with spasticity when compared to non-spastic paralyzed muscle of SCI individuals. Although most spasms were initiated at low FNS intensities, the occurrence was not consistent for all subjects and was usually diminished with repeated contractions. These findings along with the variability found in repeated measurements on two subjects show the complexity of spastic muscle response and present special problems for FNS control systems. To accommodate for these irregular contraction responses, a rapidly adapting FNS control system would be required which incorporates continuous force feedback from the stimulated muscles. Additional investigations of electrode placement, temperature, medication, skin impedance, and muscle fatigue state effects are needed.

ACKNOWLEDGEMENTS

This study was supported by the Rehabilitation Research and Development Service of the U.S. Dept. of Veterans Affairs.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

A COMPARATIVE STUDY OF ANKLE AND KNEE MOVEMENT IN PATIENTS WITH SPASTICITY AND RIGIDITY

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Measurements of spasticity and rigidity have been considered to be important in practice of rehabilitation medicine and challenged by many investigators. There are several quantitative methods for measurements of those abnormal muscle tonus.

We have been measuring ankle and knee movements by a three dimensional accelerometer for the measurements of those abnormal mluscle tonus. The accelerometer measures acceleration of the movements[1)].

Measurement of ankle movements

We measured ankle movements after single tibial nerve stimulation. The tibial nerve stimulation was used to move the ankle to plantar flexion. Then, the ankle went dorsiflexion by gravity, which stretched the triceps muscle so the triceps muscle got stretch reflex. The normal subjects showed damping oscillation. Usually 200 to 300 msec period. The spastic subjects got more oscillation. In the subjects with rigidity, it stopped suddenly with very little oscillation.

Looking at EMG of triceps surae, there was no EMG in normal case. In spastic cases, EMG looked like assisting oscillation. And in rigidity case, EMG looked like stopping ankle movements. The damping ratios were measured and were compared. We believe, at least, spasticity can be quantified by this method[2)].

A model was constructed. Movements were simulated successfully by a computer.

Measurement of knee movements

The system of measurement of knee movements was similar to Vodovnik's method[3)] except using an accelerometer in stead of a goniometer. Patients sat on the bed. The accelerometer was attached to the bars which were fixed so that the axis of accelerometer was in sagittal plane, perpendicular to the long axis of leg and 60 cm apart from the center of the knee. EMG was introduced from rectus femoris muscle with a pair of surface electrodes, 8 cm apart each other at the middle of muscle belly. For the measurement, the leg was lifted passively up to 60 degrees from vertical line then let it drop freely.

Subjects were 10 patients of Parkinson's syndrome as rigidity cases but five could not relax at the examination and were excluded. Nine spinal cord injury patients and one hemiplegic patient were examined as spastic cases. Spastic cases were divided into three level as one severe, four moderate, five mild according to their knee and ankle jerks, and, ankle and patellar clonus. After two or three times of trials, the experiments were performed. The results were compared to 10 normal subjects.

Results

The results were shown on Fig.1. In the top of right, the recording of normal case was shown. In the normal, there was no EMG with leg movements, that was, the oscillation was mainly due to the passive pendulum movements with the period from 1 sec. to 1.5 sec.

In the rigidity case, at the middle, the curve was more oscillating than the normal with period from 800 to 900 msec and there were EMG at rectus femoris synchronized with leg movements.

In the spastic cases, the leg movements were somewhat different to the rigidity cases. In the mild case, the leg movements were oscillating but less comparing to the normal with EMG at rectus femoris. In the moderate case, at the bottom of Fig.1, the curve had an initial wave with big ampulitude and almost no oscillation with EMG of more ampulitude and longer duration. In the severe case, the curve had several oscillations with big ampulitudes and more frequencies, with EMG synchronizing with leg movements.

So the reflex muscle contractions were produced to suppress or to facilitate leg movements depending on the level of spasticity. These findings were compatible to Vodovnik's report.

Simulation

The diagram of total system was shown on Fig.2. In anterior horn of spinal cord, α-motor cells and γ-motor cells were divided into dynamic and static. α-motor cells received the controlling impulses (Fe, Fi) from upper motor neurons. The impulses from upper motor neurons were excitation and presynaptic inhibition to the feedback impulses. In the skeletal muscle, the impulse-contraction-conversion system contracted the muscle. With stretching of muscle, the nerve endings at intrafusal muscles were excited. Those impulses reached to the α-motor cells with certain time delay.

The knee movement was initiated by passive drop of leg and it depended upon viscosity, elasticity and enertia around the knee. All of these were expressed as muscle-link system.

The knee movements were simulated by a computer based on this diagram. For the simulation, those constants which were shown at the model, were quoted from other articles[2), 3)]. Some were measured by us.

There were six variables which were necessary for the simulation. Those were

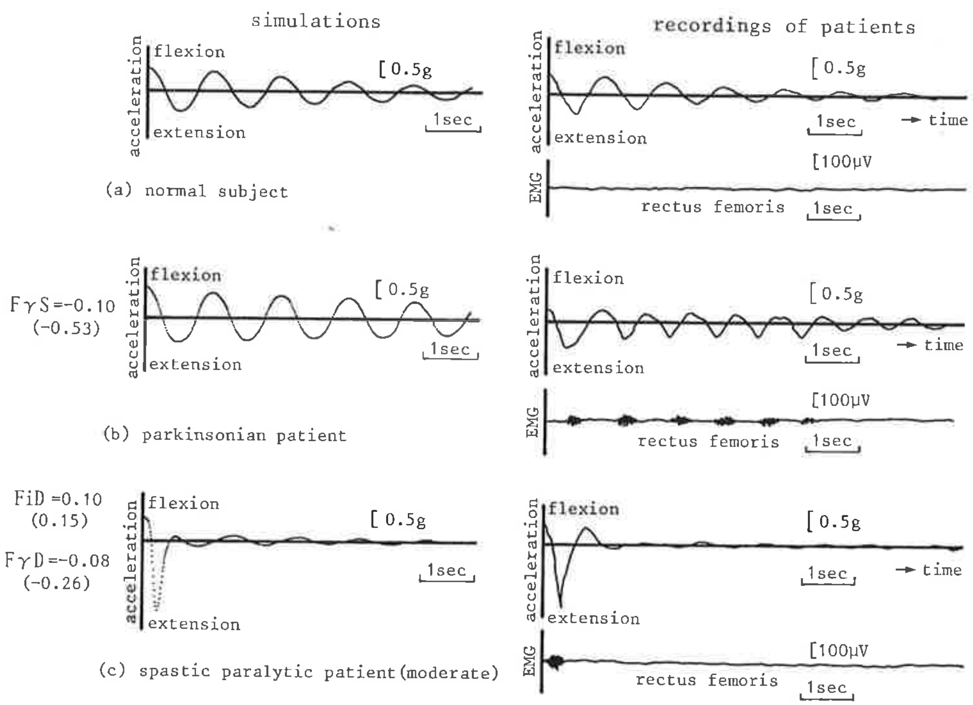


Fig. 1. direct measurements and simulations

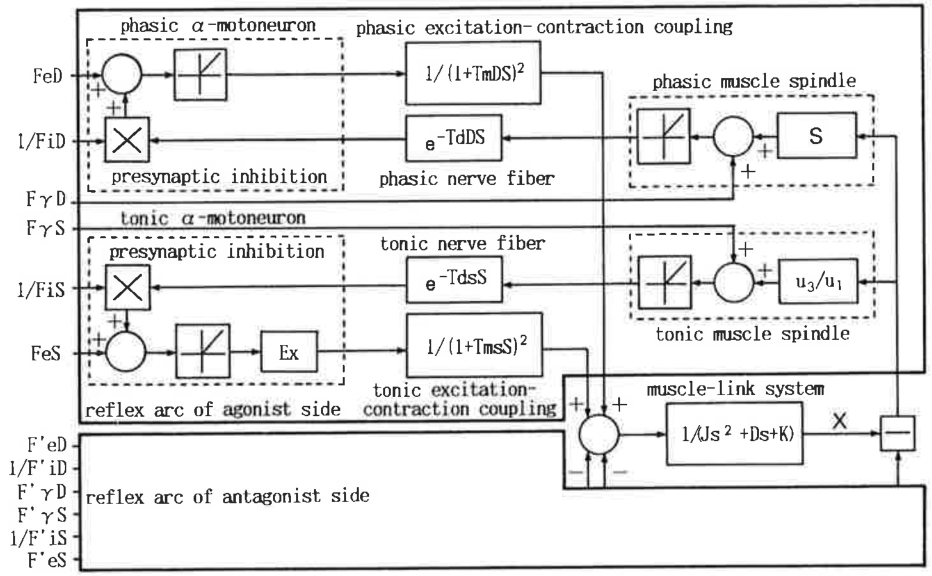
impulses from upper motor neurons, which were FeD, FeS to α-motor neurons for excitation, FiD, FiS as inhibiting impulses to feedback impulses, and, FγD, FγS as impulses to γ-motor neurons. The out put of simulation was acceleration of lower leg.

The computation gave the results shown at left side of Fig.1. Those were compared to the direct measurements.

Discussion

For the measurements of abnormal muscle tonus, many methods have been informed. Our method is considered as one of the kinematic studies. There are two important meanings in this experiment. Firstly, it is possible to evaluate the abnormal muscle tonus by means of acceleration curve itself. Secondly, it also is possible by knowing variables of diagram. Regarding the reciprocal patterns of acceleration curves of knee and ankle movements in rigidity and spasticity, it can be explained by the differences of influencing factors such as enertia, elasticity and viscosity around the knee and ankle.

Considering the measurements of rigidity, there are some problems to be solved such as getting a relaxation from the patients and positioning of the patients.



J: moment of inertia around the knee D: viscosity in knee K: elasticity in knee
s: laplacian u_3/u_1 : ratio of sensitivity of velocity to that of length in spindle
Td: sum of propagation delay in time of GIa fiber and α-fiber
Tm: twitch contraction time x: length of muscle

Fig. 2. diagram of simulation

Conclusion

1. However the time scales were different, the acceleration curves of knee movements of rigidity patients were very oscillating and were similar to that of ankle movements of spastic patient.
2. The acceleration curves of knee movements of mild and moderated spastic patients had very little oscillation and were similar to that of ankle movements of the patients with rigidity with period of one fifth in the time scale.
3. The model was constructed. By that model, the mechanism of spasticity and rigidity can be explained as far as movements of ankle and leg concern.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

EMG AND HEART RATE PATTERNS SUBSEQUENT TO BRIEF TREADMILL WALKING

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INTRODUCTION

Cerebrovascular accident (CVA) is a serious medical problem in modern society. Although not necessarily fatal, CVA can result in physical disability because of the impairment of funtional movement (1). One of the most common problems is due to loss of inhibition to antagonistic muscles. This means that instead of relaxing, these antagonastic muscles display excess tone or spasticity and interfere with the movements attempted by agonistic muscles (2). Several different therapeutic exercise approaches are currently used in the Physical Therapy profession to retrain the CVA patient's muscles to allow for smooth limb movement (3,4,5).

One approach that is used to reduce muscle spasticity is biofeedback training (BT). The electric activity (EMG) of the spastic muscle is picked up by surface mounted electrodes, amplified and displayed for the patient as either visual or auditory stimuli (6,7). Unfortunately, improvements seen with BT often do not persist after the patient is removed from the instrument. When a forceful agonistic muscle contraction is later attempted, high levels of unwanted tone may return to the antagonistic muscle (8,9,). For long term effectiveness the positive effects of BT must be prolonged beyond the direct use of the biofeedback instrument.

We propose the following research question: to what extent will physical activity perturb the relaxation state as indicated by EMG biofeedback and heart rate? The purpose of this study was to determine if normal subjects could regain a state of relaxation after a brief exercise bout consisting of walking on a treadmill.

METHODS

Subjects

Ten normal, healthy subject (M=3, F=7) without neurological impairment volunteered for this study. Age ranged from 21-53 years. All subjects signed a document of informed consent.

Equipment

EMG. Electromyographic data were collected by J and J EMG models M-52 and M-53 set to 1 microvolt full-scale range. Averaging time was set to .1 sec. A J and J Digital Integrator Model D-200 set to a 2 sec time base and a 0 sec delay time was used to display digital data for recording purposes

One cm diameter recessed surface electrodes, snap leads, double adhesive collars, and commercial electrolyte gel were used for surface EMG contact.

Treadmill. Walking was performed on an Invacare Ergometer Motorized Treadmill Model FT 2400.

Heart Rate. An Invacare Heart Rate Monitor Model FT 4200 with a light beam ear clip was used.

Recliner. A hospital grade geriatric chair was used for subject rest intervals. While the chair could achieve 6 different positions, the second reclined position was used for this study. The chair included a foot rest and head rest. Castors provided ease of movement, and side shelves were used to hold equipment.

Procedure

Two sets of surface electrodes were attached to the subjects. The first set for the left trapezius was midway between C6 and the acromion with reference contact over the acromion. The second set for the frontalis was at the hairline with the reference contact anterior to the other two electrodes. The heart rate (HR) monitor ear clip was attached to the left ear lobe.

Subjects rested for 10 minutes in a comfortable reclined chair to establish baseline EMG and HR values. After the rest period the subject walked on the treadmill for 5 minutes at 3 mph (4.8 kph) and at a 3% grade. Only HR was monitored during the walk. The subject then returned to the recliner chair. EMG and HR were recorded for immediate recovery (within 30 sec) and then at 2,4,6,8, and 10 minutes after cessation of walking.

Analysis

Data from all subjects were combined and submitted to one-way ANOVA's using "time" as a repeated variable. Three different ANOVA's were performed using trapezius EMG, frontalis EMG, and HR as dependent variables.

RESULTS

A summary of results is presented in the table.

TABLE
EMG AND HEART RATE AFTER TREADMILL WALKING
( ) = standard error of Mean

Time(min)	Frontalis EMG (uv)	Trapezius EMG (uv)	Heart Rate
Rest	.603 (.060)	.671 (.068)	71.1 (3.7)
.5	.489 (.098)	.762 (.135)	83.1 (5.7)
2	.392 (.075)	.584 (.060)	79.3 (5.7)
4	.427 (.028)	.558 (.049)	78.3 (5.6)
6	.530 (.062)	.569 (.056)	77.0 (5.4)
8	.481 (.063)	.768 (.160)	75.6 (5.4)
10	.366 (.058)	.628 (.075)	76.8 (5.3)

None of the three variables produced significant F values.

DISCUSSION

These subjects appear to easily reattain a state of relaxation within a few minutes of mild exercise. Although most active individuals would probably agree that the time to recovery would be longer after more vigorous activitiy, the effect of moderate or heavy exercise on relaxation is not predictable from these results.

A more important concern is that of whether BT can be used to shorten the time to relaxation after physical activity.

To test this premise on normal subjects apparently would require a more intense exercise bout. The subjects could then be tested before and after a number of BT sessions.

Further work should use subjects with spasticity in isolated muscles. These muscles could be monitored for EMG activity before, during and after exercise. These same subjects could then receive a few weeks of BT and be re-tested.

SUMMARY

Non-neurologically impaired subjects can reattain a relaxed state after brief, mild exercise as indicated by frontalis EMG, trapezius EMG, and heart rate. This relaxed state can be achieved in less than 10 minutes after the cessation of activity.

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APPLIED MOTION ANALYSIS

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

MULTIFACTORIAL ANALYSIS OF COMPLEX NATURAL MOVEMENTS

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INTRODUCTION

Basic studies on motor control have been mainly focused on invertebrates or lower mammals, most often able to execute very specialized performances, sometime quite skilled, but with a high degree of stereotypia. When the experimental investigation and the mathematical modeling were directed to primates and humans, descriptions have mainly considered reflex responses or simple voluntary movements involving one or two joints, the rest of the body being usually constrained or anyway neglected.

On the other hand, applied studies of motor analysis are usually dealing with natural movements performed by "freely behaving" unconstrained subjects. Indeed, in this kind of movements a distinctive feature of the motor system is maximally exploited, namely the possibility to execute a same motor task through different combinations of muscle forces and/or kinematics (different strategies). Such an operational redundancy, which implies a peculiar organization of the control mechanisms exceeding the classical concepts of control theory, will optimally adapt the resulting movement to the specific individual and environmental demand (1, 2, 3). The analysis of these aspects, as well as the identification of the various strategies adopted in complex movements requires a rigorous investigation of different kinds of variables including kinematics, forces and EMG. Starting from these data, suitable mathematical models are available for the computation of other parameters which cannot be measured directly (e.g. mechanical moments acting at the joints, muscle lengths, etc.).

TECHNOLOGICAL DEVELOPMENTS

In recent years improved technologies are available which make it possible quantitative analysis of movement in a more rapid and reliable way, with minimal interference on the subject's performance.

The ELITE system (4), which has been recently developed using criteria from the computer vision technology, exhibits peculiar features which make it suitable for the analysis of kinematics of voluntary unconstrained movements.

Small hemispherical passive markers are placed on the relevant points of the body, and two or four TV cameras take the subject during movement. Two levels of intelligence characterize the processing system. The first level includes a specially designed processor that is the core of the system and uses very fast VLSI chips arranged in a parallel architecture. This level provides in real time a bidimensional cross-correlation processing of the TV signal, so that the markers are automatically recognized only if their shape matches a predetermined "mask". The procedure allows for a great reliability of the marker detection and for a high accuracy in the computation of the coordinates.

The second level of intelligence is implemented on a general purpose computer (host computer). Basic operations as kinematic data enhancement, tracking and reconstruction of the hidden markers, correction of distortion and three dimensional (3D) reconstruction by stereometric techniques belong to this level.

A second stage of the model receives additional input from the force plate (which measures the ground reactions), and computes a set of variables related to dynamics and muscular kinematics.

Surface EMGs are recorded during free movements by an 8 channels telemetric system. A preamplifier mounted on the electrode bar is used to improve the signal/noise ratio. EMG signals are digitized, preprocessed by a small portable device and transmitted in digital form to a remote receiver, directly interfaced with the general purpose computer.

INDIVIDUAL STRATEGIES

In the following two examples of how variability in the execution of natural movements can be objectively documented and quantitatively evaluated will be presented.

Gait. In a study of normal gait (3, 5) on a group of young male subjects walking on a walkaway at a natural speed, EMGs from the main lower limb muscles, kinematics and ground reactions were recorded. If we look at the EMG patterns, even at a first glance a consistent interindividual variability is clearly apparent (Fig. 1). It concerns particularly the interplay between the biarticular hamstrings (extensors at hip and flexor at knee) and the monoarticular vasti (extensors at knee).

In subject A the activity of hamstrings starts immediately before the heel strike, increases immediately there-after and lasts up to the middle of the stance phase (ST). At variance, in subjects B and C the hamstring activity reaches its maximum at, or before the heel strike, and stops immediately after. The vasti are slightly activated in subject A, while displaying an increasing level of activity in subject B and, particularly strong and long lasting in ST, in subject C. This subject reveals also a bimodal activity of the calf muscles, which is not present in subjects A and B.

A question arises at this point: can these differences in amplitude and timing of EMG patterns be associated with other independent variables so that they can be reasonably interpretated as a component of a more complex individual motor strategy? The comparison with the moment time course computed independently from kinematics and ground reaction measurements confirms that the different EMG patterns are the spy of a well defined and interindividually variable dynamic behaviour. In fact, in subject A a hip extensor and knee flexor moment pattern start at the end of swing phase and last for almost the entire STance, in both cases supported by the long lasting activity of the hamstrings, the second part of the flexor moment at knee being supported by the calf muscles. In subject B, and mainly in C, the knee extensor moment is significatively higher and therefore the hamstrings stop their activity immediately after the onset of the stance, in conjunction with a more pronounced activity of Vasti. Also the bimodal time course of moment at ankle is associated with the bimodal calf activity in subject C. This close matching between individual

EMG patterns and mechanical variables, observed in each subject, excludes that interindividual differences in amplitude and timing of muscles activities could be attributed to random variability around a basic pattern, suggesting the existence of different individual motor strategies. As mentioned above a previous study (6) these individual patterns in level walking are consistent with a minimisation of the total muscle effort.

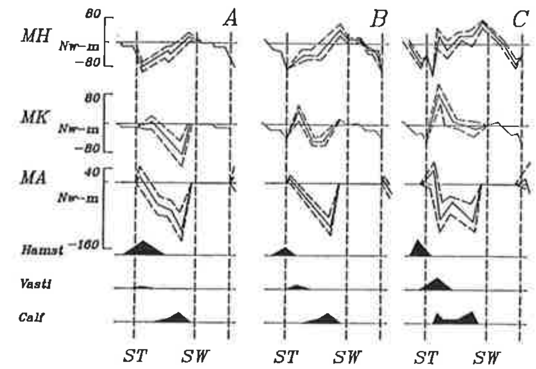

Fig. 1. Multifactorial analysis of a gait cycle in three subjects: A, B, and C. On each column from top to bottom are reported the mechanical moments active on the plane of progression at hip (MH), knee (MK) and ankle (MA) joint, respectively. The positive sign represents a flexor action at the hip, an extensor action at the knee and a dorsal flexion at the ankle. Mean values of the moments and the range of uncertainty are illustrated by the continuous and dashed line, respectively. For sake of simplicity, the activity of the main muscles has been grouped in Hamstrings (Hamst) including BFCL, ST and SM, Vasti and Calf including SO and GA. EMG envelopes are a schematization of the integral, filtered activity, normalized with respect to the maximal voluntary contraction.

Axial movements. In a study on axial movements performed while standing, the EMG pattern of trunk and lower limb muscles was investigated in association with trajectories of the main body segments and ground reactions (7). During fast backward bending of the upper trunk, the muscle groups involved are all located in the back of the body and include erectors spinae as prime movers (main target of the voluntary command), and hamstrings and calf muscles as synsergists.

Analysis of timing of recruitment of postural leg muscles (Hamstring and Calf muscles) with respect to the prime mover (Er. Sp.) revealed a rather variable relationship. In the most common pattern, recruitment of synergistic leg muscles was synchronous or slightly delayed with respect to the prime mover (Fig. 2A). In a second subgroup of subjects examined which consisted of girls with high level training in gymnastics, a consistent anticipation of calf muscles and most often of the hamstrings, was observed with respect to the prime mover (distal anticipated pattern) (fig. 2B). A most obvious question in this context is whether the distal anticipated pattern of gymnasts is actually associated with a higher level of motor performance.

Accordingly, relevant parameters related to speed of movement (peak velocity of shoulder marker), and to optimal distribution of body masses (maximal horizontal displacement of the CG) were determined over a certain range of movement excursion. Interestingly, trained subjects proved to perform the bending movement faster and with a lower displacement of CG as compared with controls, as confirmed by the statistically significant decrease of a "Performance index", represented by the ratio Maximum horizontal displacement of CG/mean velocity of shoulder displacement (37 in the gymnasts vs 73.4 in the untrained subjects).

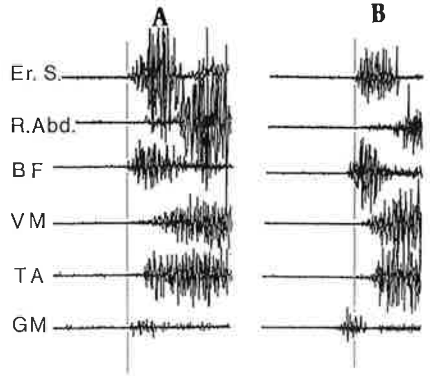

Fig. 2. Typical EMG pattern during fast backward bending of the trunk in an untrained (A) and trained (B) subject. Note the difference between the "non anticipated pattern" (untrained, left), and the "distally anticipated pattern" (trained, right). Three traces superimposed in each row. Er Sp: erectores spinae, R Abd: rectus abdominis, BF: bicepts femoris (lateral hamstrings), VM: vastus medialis, TA: tibialis anterior, GM: gastrocnemius medialis. Calibrations: 100 ms./μV. The vertical dotted line corresponds to the onset of prime mover activation (Er Sp.).

CONCLUSIONS

One of the main findings which emerges from the data presented is the existence of well defined individual patterns of muscle recruitment associated with relevant biomechanical variables in the execution of the same motor task, in the same experimental condition. The observed variables (several muscles, mechanical moments, trajectories, CG displacements) constitute only a relatively small subset of the high number of factors involved in such complex movements. However, they indicate the presence of individual motor strategies, as general schemes of organisation of a motor action, which result in the prevalence of a criterion of optimisation , including velocity, force, equilibrium, precision, energy expenditure, elegance, in order to obtain a given goal.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

AN ELECTROMYOGRAPHICAL ANALYSIS OF FAST-MEDIUM BOWLING IN CRICKET

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INTRODUCTION

The objectives of this study were to determine the sequential and temporal patterns of muscular activity during the fast-medium bowling action and to discover any intra and inter bowler differences in these patterns.

METHODOLOGY

Subjects

The subjects used in this investigation were four college bowlers with a mean age of 21.5 years (S.D. = 0.58 years) and a mean ball release (BR) speed of 29.3m/s (S.D. = 0.63m/s).

Apparatus

Electromyographical signals were monitored during bowling by an M.I.E. Medical Research Ltd. MT8 Biological Telemetry System. Raw electromyograms (EMGs) were recorded by an SC oscillograph 3006/DL with a paper speed of 125mm/s.

A Redlake Locam model 51 high speed 16mm cine camera running at 100Hz was used to film the bowler's run-up (R-U), pre-delivery stride (Pre-DS), delivery stride (DS) and follow-through (F-T).

To enable the EMGs to be temporally related to their respective film record a 'synchronisation box' was constructed and connected to the oscillograph and a Griffin Xeron stroboscope, placed in the field of view of the cine camera. Manual activation of the 'synchronisation box' illuminated the stroboscope and started the oscillograph paper rolling thus enabling the EMGs and cine film to be synchronised.

Experimental Procedure

Due to the importance of the movements of the humerus and trunk in the bowling action it was decided to analyse the muscles which control their actions. From the glenohumeral joint the *anterior* and *posterior* portions of the *deltoid* (AD and PD), *sternal portion of the pectoralis major* (PM(st)), *latissimus dorsi* (LD), *teres major* (TMaj.) and *infraspinatus* (I) were analysed. From the trunk bilateral activity in *erector spinae* (ES), *rectus abdominis* (RS) and *external oblique* (EO) was recorded.

The muscles listed above were located using extended living anatomy as recommended by Rozendal and Meijer (1). Electrodes were located on the

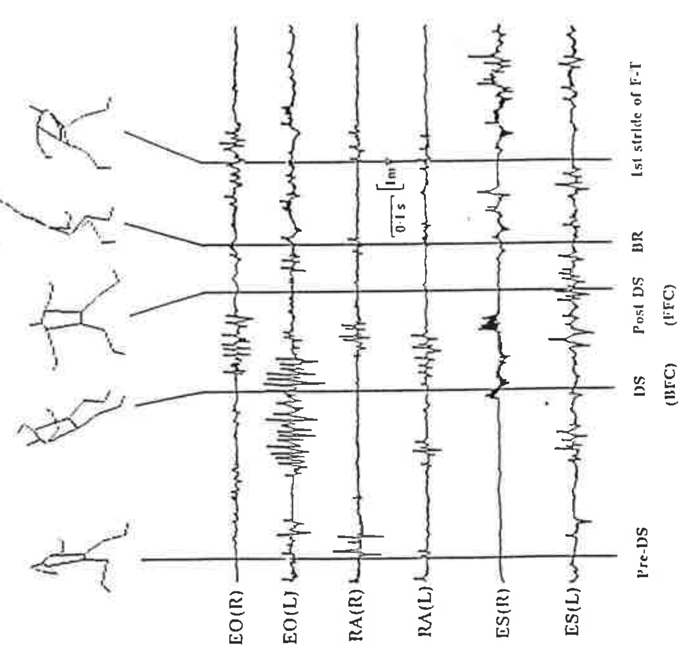

Figure 1: Electromyograms, from Selected Trunk Muscles, and Stick Figures from a Ball Delivered by GS.

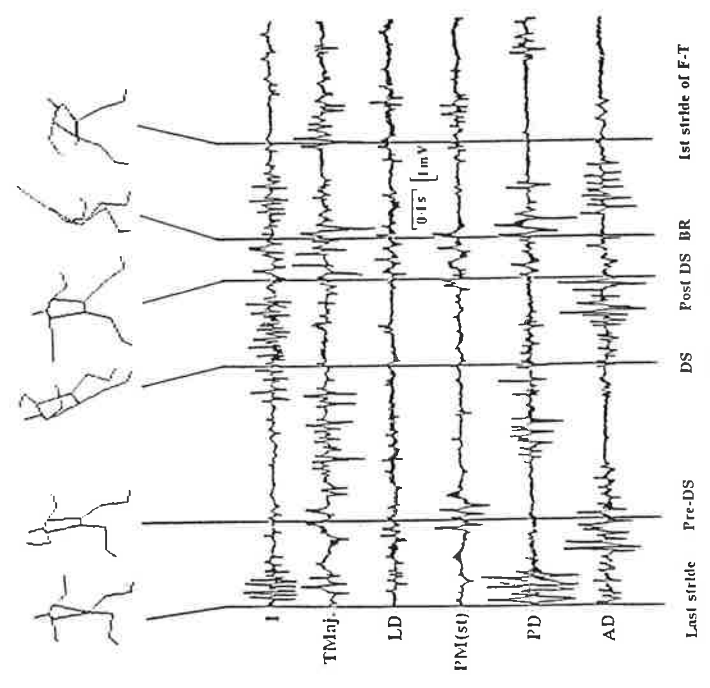

Figure 2: Electromyograms, from Selected Glenohumeral Joint Muscles, and Stick Figures from a Ball Delivered by GS.

skin overlying the *anterior deltoid* and *erector spinae* in accordance with the recommendations of Zipp (2). Electrodes overlying the *pectoralis major* were placed over the lateral crest of the muscle with the central lead point in the same horizontal plane as the areola. Those overlying the *latissimus dorsi* were positioned over its lateral crest with the central lead point in the same horizontal plane as the spinous process of the 8th thoracic vertebra. The *rectus abdominis* had its electrodes placed over the second portion of the muscle, from proximal to distal, with the central lead point in the centre of the portion thereby positioning them in approximately the same horizontal plane as those situated over the *erector spinae*, in the lower thoracic region of the trunk. Electrodes were positioned over the *external obliques* in the same horizontal plane as the electrodes overlying the other trunk muscles analysed. With regard to the other muscles, they were small enough to have their central lead points easily located over the belly.

Before the electrodes were positioned on the skin, after shaving and degreasing it with acetone, its impedance was reduced below 10 kilohms by the method outlined by Okamoto et al (3).

Correct electrode site location was ensured by performing a series of test movements which were unique to the potential functions of each muscle, as recommended by Rozendal and Meijer (2).

To give the EMGs recorded during bowling a reference spike amplitude and frequency against which to be compared against, and to observe the effects of cross-talk a series of maximal voluntary contractions (MVCs) were performed. These MVCs took the form of maximal isometric contractions during all basic movements of the humerus and trunk.

The glenohumeral joint and trunk muscles were analysed on separate occasions during bowling with at least three synchronised trials being obtained from each muscle group.

EXPERIMENTAL DATA

Stick figures and raw EMGs, from the trunk and glenohumeral joint muscles, recorded during bowling trials of subject GS are shown in Figures 1 and 2 respectively.

DISCUSSION

Figure 1 shows that before back foot contact (BFC) in the delivery stride activity was seen in the left (L) *external oblique* to rotate the trunk to the right and hence a side-on position in preparation for delivery. Before front foot contact (FFC) this activity subsided and the trunk was rotated in the opposite direction towards a front-on, ball release position by the right (R) *external oblique*. Concurrent with the latter trunk rotation activity in the *rectus abdominis* muscles initiated trunk flexion before ball release. This concentric activity occurred for GS, and one other subject, in the presence of an eccentric contraction of the *erector spinae* muscles. Following ball release the motion of the trunk was decelerated by moderate activity in all the muscles analysed.

Before back foot contact in the delivery stride the bowling arm of all subjects displayed kinematic idiosyncrasies which were reflected by inter subject differences in the patterns of EMGs recorded. Figure 2 shows that after back foot contact the circumduction and abduction of the bowling humerus before releasing the ball was controlled for GS, and two other subjects, by sequential contractions of the *anterior deltoid* and *pectoralis major*, with the medial rotation function of these muscles being neutralised by the *infraspinatus*. The deceleration of the humerus after ball release by the *posterior deltoid* indicated a ballistic pattern between it and the *anterior deltoid* for GS and one other subject.

ACKNOWLEDGEMENT

The second author wishes to acknowledge the financial support provided by the British Sports Council to enable him to attend the ISEK Symposium.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

LOWER EXTREMITY EMG-ACTIVITY OF ATHLETES AND NON-ATHLETES IN JUMPING

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INTRODUCTION

The quantity of myoelectrical activity (EMG) before and during the eccentric phase of contact has been shown in jumping to be highly correlated with contact time, contact force and angular parameters (1) among trained athletes. It is suggested (1) that this is partly to be explained by preprogrammed patterns, which are dispatched from higher centers of the nervous system (2).

In sports the effects of training on neuronal control mechanisms are interesting and important. The present study was designed to compare the EMG characteristics of lower extremity muscles in connection with some kinetic and kinematic parameters before and during ground contact in drop jumps between highly trained athletes and controls.

MATERIAL AND METHODS

Subjects

Seven triple-jumpers of Finnish national caliber represented highly trained athletes (average record 16.05±0.29 m) and eleven students with physically active life habits but no jumping training backgrounds acted as controls.

The jumpers were on average (± S.D.) 27.6±3.6 years old, 183.9±4.5 cm tall and their weight was 75.4±5.2 kg. The respective values for the control group were 20.6±2.6 years, 178.4±5.4 cm and 74.0±7.4 kg.

Measurements

The subjects were measured during one testing session in an indoor hall. This paper deals with the results collected during drop jump drills performed with bilateral foot contacts. The subjects dropped themselves from a height of 0.8 m onto a force platform and rebounded immediately as high as possible while keeping their hands on their hips during the entire jump. From three jumps the two highest were used for further analysis.

TABLE I
KINEMATIC AND KINETIC CHARACTERISTICS

Variables	Jumpers	Controls	t	p<
Jumping height (cm)	47.3±6.7	35.3±5.2	3.98	.01
Contact time (ms)				
-eccentric	97.2±14.2	152.3±48.7	3.53	.01
-concentric	68.6±7.7	95.5±23.6	3.49	.01
Vertical force (N/1000)				
-eccentric	4.32±0.71	2.69±0.86	4.44	.001
-concentric	1.70±0.36	1.06±0.28	4.01	.01
Knee angle (degree)				
-touch down	156.9±2.3	141.0±4.5	9.89	.001
-deepest position	118.4±2.6	99.4±5.9	9.30	.001
-takeoff	173.6±3.3	165.3±9.3	2.70	.05

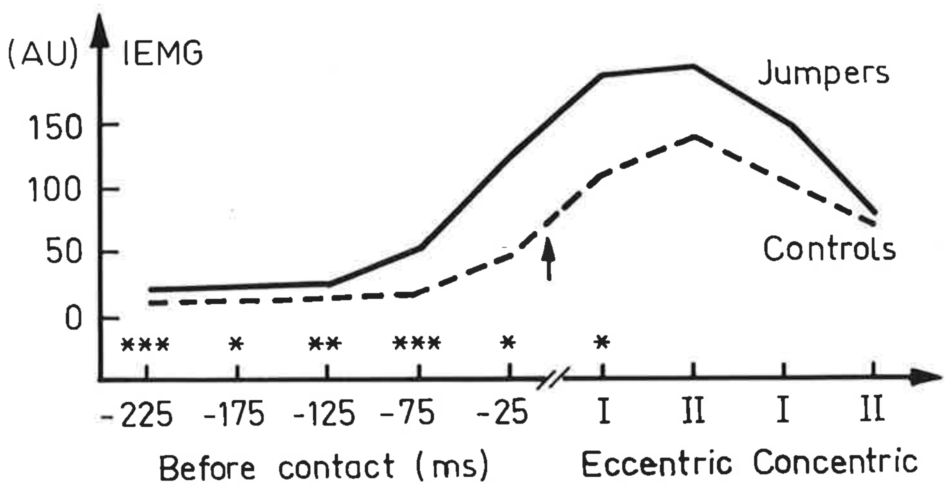

Fig 1. IEMG of the m. vastus lateralis before and during contact in arbitrary units (AU).. The arrow points to the touch down. The stars denote the statistical significances of differences between the groups. * p<.05 ** p<.01 *** p<.001

Data collection

EMG-activity was registered with bipolar surface electrodes (Beckman) on the muscle belly of m. vastus lateralis, rectus femoris, biceps femoris and gastrocnemius. The preamplified EMG signals were telemetrically (Biomes 2000) transmitted to a magnetic tape recorder (Raçal) simultaneously with three dimensional ground reaction forces from a force platform. The jumps were filmed with a Canon Scoopic camera (64 fr/s).

Data analysis

The EMGs were full wave rectified and integrated for 50 ms periods during 250 ms prior to ground contact, and for the first and second halves of the eccentric as well as concentric phases of contact. The separation between the eccentric and concentric phases of contact was based on the knee angular displacement. The flight times of the jumps were used to calculate the height of rise of the body's center of gravity. Eccentric and concentric contact times, and average forces were calculated from the force-time curves. Knee angles were analyzed (a Summagraphics digitizer) from the film for the touch down and takeoff phases as well as for the deepest bending position (between eccentric and concentric phases). Statistical calculations included mean, standard deviation of the mean and Student's t-test.

RESULTS

The triple-jumpers jumped 34 % higher, their total contact time was 33 % shorter, average eccentric force 62 % and concentric force 60 % greater than the respective values of the control group. The jumpers had greater knee angle values at touch down, at the deepest position as well as at take-off than the controls. All the differences between the two groups were statistically significant as shown in table 1.

The EMG activities of the m. vastus lateralis were statisticallly significantly greater for the jumpers before ground contact and during the first half of the eccentric phase of contact than for the controls (fig. 1). EMGs measured from the m. gastrocnemius and m. biceps femoris during the first half of the eccentric phase were also greater (p<.05) for the jumpers than for the controls, while in respect of the other time intervals as well as for the m. rectus femoris the differences between the two groups did not reveal statistically significant differences.

DISCUSSION

The touch down velocity of the two subject group was equalized by dropping all the subjects from the same height. The triple-jumpers activated their m. vastus lateralis before contact more than the control group. This may mean that the groups differed in respect of their "preprograms". Preactivation was found in our previous study (1) to be highly correlated (r=.98) with eccentric EMG activity. During the eccentric phase of contact high muscular activity due to high preactivation, and reflex potentiation in connection with high voluntary activity favor storage of elastic energy, which in turn during the concentric phase of contact can be utilized for high power production (see 1). The differences in EMG activities between the groups indicate that the neuromuscular system of the jumpers as compared with that of the non-jumpers was better prepared to withstand and utilize strong eccentric stretching. Following these the eccentric contact time was shorter and eccentric force greater than among the controls. Better utilization of the eccentric stretching phase was seen in greater force production during a shorter time in the concentric phase, and in the higher jumping height. Interestingly the jumpers' knee angles were greater than those of the controls, which indicates that the jumpers most probably worked with shorter muscle and sarcomere lengths than the non-jumpers. Whether the differences between the groups are due to differences in the inherited structure and function of their neuromuscular systems or due to different training backgrounds is an open question and a target of research in the future.

ACKNOWLEDGEMENTS

This study was supported by the Finnish Central Sports Federation. Mr Ossi Aura (Lic.Phil.), Mikko Levola (MSc.), Jarmo Mäkelä (MSc.) and Ari Pelander as well as the Kuortane Sports Institute are thanked for their co-operation and help in the collection od data.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

GROUND REACTION FORCES AS BIOMECHANICAL MEASURES OF EXCESSIVE PRONATION

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INTRODUCTION

Running has taken a leading role among mass-participation sports, with an estimated 30 million runners in the United States alone. Unfortunately, approximately five million runners annually sustain injuries that involve the musculoskeletal system. Up to 60% of these injuries are related to excessive pronation.

Pronation is a triplanar motion of the foot which includes eversion, abduction and dorsiflexion (Vogelbach and Combs, 1987). To a certain degree pronation is beneficial because it acts as a shock absorbing mechanism, and allows the foot to adapt to an uneven terrain (Clarke et al., 1983). Excessive pronation can be divided into two categories: acquired and congenital. The former follows trauma or a systemic disease. The latter is a dynamic instability caused by soft tissue or bony anomalies. Excessive pronation causes an abnormal delay in the onset of resupination during the latter part of stance (Pagliano, 1987). Resupination is the portion of the step where the center of mass of the body is moving forward and laterally and the foot begins to move upward preparing to leave the ground. Improper resupination impairs the foot's capacity to stabilize and act as a rigid lever, resulting in a less efficient and less powerful propulsive phase. Excessive pronation also causes a collapse of the medial longitudinal arch of the foot during the weight bearing phase of gait. As the elastic properties of the arch of the foot play an important role in the energy storage and return, and reduce the amount of work which must be done by the muscles during running (Ker et al., 1987), the collapse of the arch could lead to a less efficient gait.

Orthotic devices, if properly prescribed and constructed, can have an impact on decreasing injuries associated with excessive pronation. Ideally, an orthotic device controls motions that are abnormal or undesirable, and permits motion where normal function can occur (AAOS, 1985). For orthotics to achieve optimal results they need to be evaluated both subjectively by the runner and objectively using biomechanical tests. The objective of this study is to present

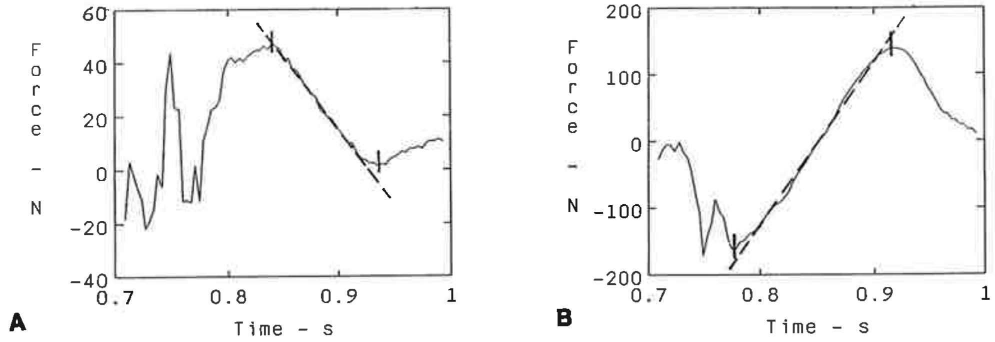

Figure 1: A. The mediolateral resupination slope. B. The anteroposterior propulsive slope

some biomechanical measures that can be used to achieve accurate diagnostic procedures and to objectively evaluate the efficacy of orthotics.

METHODOLOGY

An experiment was designed to investigate the ground reaction forces generated during running for both a control group and a group of excessive pronators. The subjects from the latter group were diagnosed as excessive pronators following a clinical examination and orthotic shoe inserts were individually prescribed and constructed for them. An AMTI force plate, embedded in the track over which the subjects ran, was used to measure and record the three moments and the three interactive forces between the foot and ground. The mediolateral, anteroposterior and vertical ground reaction forces were plotted as functions of time. The following parameters were extracted from the data:

- Mediolateral Resupination Slope - the first order regression slope of the resupination phase of the mediolateral force trajectory, normalized by body weight. The resupination phase is defined as the portion between the point which begins a final decrease in medial force magnitude and the point where this slope levels off (fig. 1).

- Anteroposterior Propulsive Slope - the first order regression slope of the anteroposterior propulsive phase between the maximum braking (decelerative) force and the maximum propulsive (accelerative) force, normalized by body weight (fig. 1).

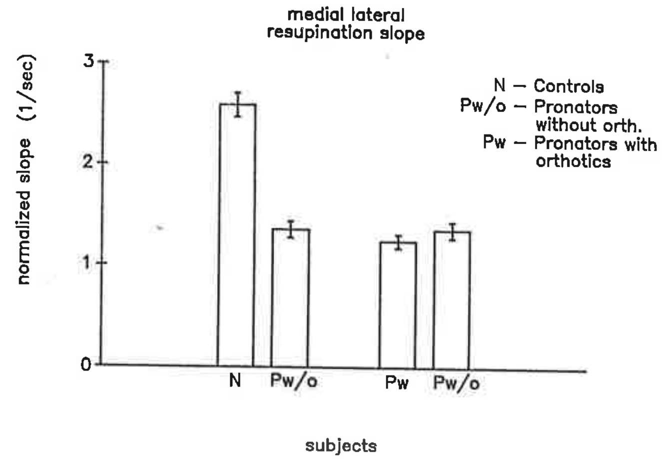

Figure 2: Mediolateral resupination slope differences

RESULTS AND DISCUSSION

Figures 2 and 3 present the differences in the magnitude of the mediolateral and anteroposterior slopes of the control group and the pronated runners (with and without orthotics).

Results indicated that the slope of the resupination portion of the mediolateral force trajectory was 1.75 times larger for the control group as compared to the excessively pronated group not wearing orthotics (fig. 2). By modelling the arch of the foot as a mass-spring system, this slope can be related to the spring coefficient of the arch of the foot. Thus, normal runners have approximately twice the spring coefficient of excessively pronated runners. In addition, results showed that orthotics did not increase the magnitude of the slope for excessively pronated subjects.

Analysis also indicated that the control group's mean anteroposterior propulsive slope was 20% greater than that of the excessive pronators (fig. 3). This could be related to the fact that excessive pronation causes improper resupination, which in turn causes the foot to remain flexible during toe-off resulting in a less efficient propulsive phase. Orthotics did not increase the propulsive slope and therefore did not increase the efficiency of propulsion. All results were significant to the 90% confidence level.

CONCLUSION

The mediolateral resupination slope and the anteroposterior propulsive slope have emerged from the experimental results as biomechanical measures of excessive pronation. In addition, these parameters were useful in the evaluation of the efficacy of orthotics. In the future, these results could be used to enhance diagnostic and prescriptive procedures for orthotics.

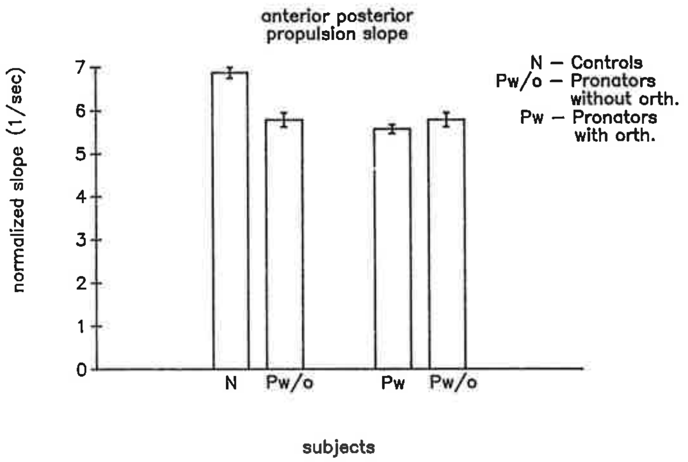

Figure 3: Anteroposterior propulsive slope differences

ACKNOWLEDGEMENT

This research was supported by Reebok International, Ltd. The authors acknowledge original contributions by Evan Sherr and Joe Heally.

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Electromyographical Kinesiology
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ELECTROMYOGRAPHIC ANALYSIS OF THE WALKING GAIT CYCLE

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INTRODUCTION

There has been some question as to the validity of intersubject Integrated EMG (IEMG) averages, as applied to the walking gait cycle, when the IEMG Reference Levels (RL) were derived from isometric Maximum Voluntary Contractions (MVC), as in earlier studies, as by Dubo et al. (1), Arsenault et al. (2), or from a 50% MVC, as in the study reported by Yang and Winter (3), although in the latter study, and in that by Winter and Yack (4) much reduced variability was found, when intrasubject ensemble stride averages were used as the normalization factors. In these earlier studies the normalized Relative IEMG Amplitudes (RA) were determined by dividing the Absolute EMG values at points in the stride cycle by mean stride EMG values, and then determining the Coefficients of Variation (CV) for these points across the entire stride cycle. The objective of this present study was to reexamine the variability, and, therefore, the validity of intersubject ensemble averages by determining the average RA and Standard Deviation (SD) in segments (or phases) rather than at points in the stride cycle.

MATERIALS AND METHODS

A separate RL of surface EMG was established for the Vastus Lateralis (VL), Medial Head of Gastrocnemius (MHG) and Soleus (S) muscles in two female and three male subjects by averaging three 25% isometric Maximum Voluntary Contractions (.25 MVC) performed on a special apparatus which positioned a resisting strap in series with a strain gauge, being monitored on an oscilloscope visible to the subject (Figure 1), while rectified EMG, averaged at 100 msec intervals, referred to here as IEMG, was simultaneously being digitized at 100/s on a Model 6300 AT+T computer. Immediately after the RLs were established, the subject walked on a level constant speed treadmill (4.3 KpH), while two 40 second samples of the electrical output of footswitches and the IEMG were being digitized and stored for further processing.

Twenty strides were manually selected from the 80 seconds of recording, normalized to percentages on the time axis, and the absolute EMG amplitudes at each 1% interval normalized to a RA in two ways: 1) by dividing the absolute IEMG amplitude by the RL (IEMG/.25 MVC); and 2) by dividing the absolute IEMG amplitude by the Average IEMG for the entire stride cycle (IEMG/AveIEMG).

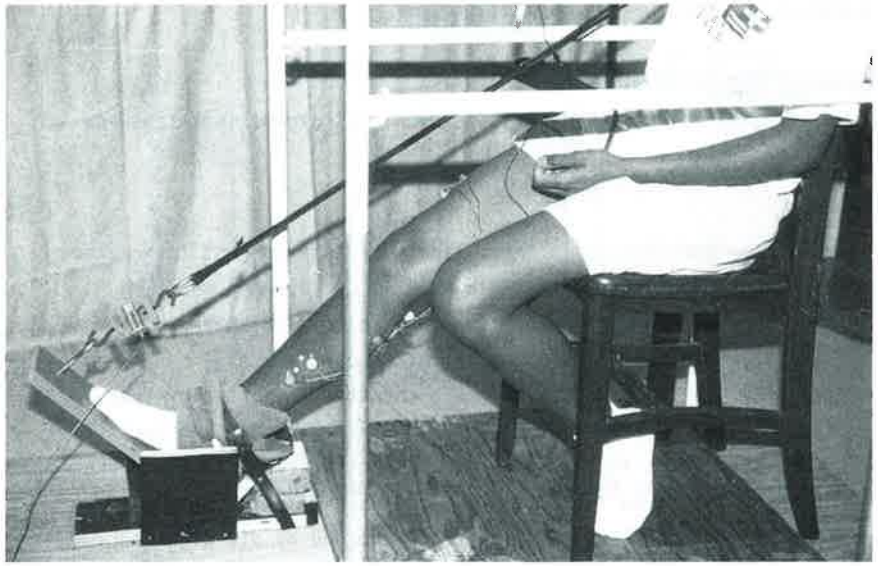

FIG.1 IEMG REFERENCE LEVEL DETERMINATION FOR MEDIAL HEAD OF GASTROCNEMIUS

Intersubject ensemble averages were derived from the one hundred strides used to determine the five intrasubject ensemble averages.

SDs were determined by two methods: 1) at 20 points, which corresponded to 5% intervals along the course of the stride, and 2) for the average amplitudes of five Foot Switch Phases: 1) Swing Phase, 2) Heel Contact; 3) Heel and Ball (Foot Flat); 4) Ball (Heel Rise); and 5) Toe Contact. The Coefficient of Variation (CV) was determined by dividing the average SD by the average Relative IEMG amplitude across the stride.

RESULTS

The variability of Relative IEMG Amplitudes as expressed by the CV, was about twice as high, when using the .25 MVC Reference Level as when using the AveIEMG, for the normalization factor (F Ratio = 27.41, p = .000 for ANOVA), but little difference between the CVs for the modes of SD determination (F Ratio = 0.39, p = 0.540 for ANOVA), except that for the S the Foot Switch Phase

amplitudes were conspicuously less variable than those at the 5% points.

When the Ave. IEMG was used as the normalization factor, differences could be seen in the consistency of the Relative IEMG Amplitudes among the five Foot Switch Phases of the gait stride cycle. Table 1 summarizes the intersubject variability of the Relative IEMG Amplitude ratios between the two muscle groups (Ratio 1:VL/Ave.of MHG+S) and between the two calf muscles (Ratio 2:MHG/S) for the five Foot Switch Phases.

TABLE 1

INTERSUBJECT VARIABILITY OF AMPLITUDE RATIOS
(NORMALIZATION BY IEMG/AVE IEMG)

Foot Switch Phases	1	2	3	4	5
Ratio Averages	(1) 1.44 (2) 1.22	(1) 3.22 (2) 0.73	(1) 0.81 (2) 1.13	(1) 0.27 (2) 0.65	(1) 0.49 (2) 0.71
SD	0.49 0.59	0.58 0.28	0.35 0.39	0.06 0.06	0.16 0.12
CV	0.34 0.48	0.18* 0.38	0.43 0.35	0.22* 0.09*	0.33 0.17*

(1): VL/MHG+S; (2): MHG/S * Low variability at weight acceptance (Phase 2) and forward thrust (Phases 4 and 5).

DISCUSSION

Segmenting the EMG amplitudes into footswitch phases of the stride cycle provided a means of relating relative muscle amplitudes, and intersubject variabilities to muscle function, and a partial answer to the query of Arsenault et al. (2) as to whether criteria for "normal" gait exist. During Heel Contact and Heel Rise (Phases 2 and 4), when strong force vectors were applied for accepting weight and the forward thrust, respectively, the RAs of the Vastus and Calf muscle IEMG were quite consistent among the five subjects. In contrast, in Foot Flat (Phase 3) which was functionally a transitional period, when the body weight was being transferred from the heel to the ball of the foot, the RAs of these two muscle groups were much less consistent. When .25 MVC was used as the normalization factor these ratios showed high CVs for all the Foot Switch Phases.

CONCLUSIONS

1. THE COEFFICIENT OF VARIABILITY (CV) was significantly LESS when Relative IEMG Amplitudes were determined by IEMG/AVE. IEMG rather than by

IEMG/.25 MVC, in agreement with Yang and Winter ('84).

2. DIFFERENCE IN IEMG VARIABILITIES (CVs) were detectable in different gait cycle phases USING IEMG/AVE. IEMG, but NOT IEMG/.25 MVC.
3. VARIABILITIES (CVs) of IEMG AMPLITUDE RATIOS for VL and MHG were the LOWEST in the WEIGHT ACCEPTANCE (Heel Strike) and FORWARD THRUST (Heel Rise and Toe Off) gait cycle PHASES.
4. The consistency of the IEMG AMPLITUDE RATIOS between the vasti and calf muscles AT the HEEL STRIKE and HEEL RISE PHASES SUGGESTS THEIR USE AS DIAGNOSTIC CRITERIA FOR NORMAL GAIT.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

REDUNDANCY RESOLUTION IN A PROSTHETIC ARM USING A KINEMATIC CONSTRAINT

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ABSTRACT

Controlling a four degree of freedom powered arm prosthesis by body movements on a one-to-one basis places substantial mental burden on the amputee. In terms of feedback to the amputee, body controlled prostheses are considered by some to be superior [1], compared with mioelectric controlled limbs. To this end, multifunction arm prostheses must be controlled in an integrated manner. The present work is an investigation into the control methodology for shoulder disarticulated amputees, where the activation signals are three displacements gathered from residual motion of the shoulder. This paper outlines three possible constraints (considered from a robotic and biological standpoint) employed by the physiological system in eliminating redundancy in performing optimal and 'elegant' motion. A comparison between methods employing these geometric constraints and actual data is made.

INTRODUCTION

In order to emulate the natural arm, a shoulder disarticulated amputee requires a multifunctional prosthesis with at least four degrees of freedom for positioning the hand in space (flexion-extension of the upper arm, ϕ; abduction-adduction of the upper arm, θ; humeral rotation, ψ; and elbow flexion, α). Controlling a multifunction arm such as the one just described on a one-to-one basis (one signal for one corresponding degree of freedom) would place considerable mental strain on the amputee. At this level of amputation it becomes imperative that an arm is controlled in a coordinated way where all the degrees of freedom are active simultaneously with minimal cognitive coordination by the amputee.

The use of body movements as control signals is deemed superior because it utilizes the natural systems proprioceptive properties to provide a degree of awareness of arm position and velocity that is essential to the amputee. The awareness is much less when EMG control is used. Simpson in 1964 was the first to talk about reducing mental load, providing position control and at the same time duplicating the physiological arms inherent feedback features. He defined the term 'Extended Physiological Propioception' (EPP), which is the ability to point with a rigid object such as a pointer, without needing to look at it. Simpson, Childress and others [1,2,4] provided EPP by linking the amputee and arm by direct cables that give the amputee an immediate feel for what the arm is doing.

In the present work, concepts of EPP are used in conjunction with microprocessor control to provide a control methodology for a powered arm that uses three independent displacement signals at the shoulder to navigate the arm in space.

This paper attempts to solve the problem of going from three signals to four degrees of freedom by resolving the redundancy in a manner that results in arm motion that is consistent with physiological movement from a functional and aesthetic standpoint.

METHODOLOGY

Motion of the shoulder girdle was monitored and the space envelope of the movement of the Acromion was defined [3]. Although the mediolateral aspect is limited, three displacement signals may be acquired. The mapping between shoulder displacements and hand position in 3-D space would provide the amputee with a mechanism to navigate the hand in space by moving the shoulder in the appropriate direction. The arm would act as an extension of the shoulder which would provide a degree of proprioception which depends on the reliability of the system.

The displacement of the hand in Cartesian space needs to be mapped into joint space to provide the input to the four actuators. In order to go from three displacements of the hand to four joint angles the inverse kinematics must be solved and a constraint that eliminates the redundancy introduced. The following constraints were investigated:

1. *Minimization of total joint travel* . By using the Moore-Penrose pseudoinverse the 3 x 4 Jacobian matrix, J, can be inverted to yield four joint variables,

$$[\Delta\theta] = [J]^{+} [\Delta x]$$

where $[J]^{+} = [J]^{T} ([J]\,[J]^{T})^{-1}$ is the pseudoinverse.

The implied optimality criterion here is,

$$\sum_{i=1}^{4} (\Delta\theta_i)^2 = \text{minimum}$$

The total travel of the joints is minimized. Although this criterion is applied to industrial robots, it bears consideration from a biological standpoint.

2. *Fixing upperarm-forearm plane*. From an empirical standpoint it was hypothesized that rotation of the upperarm-forearm plane, for routine motions, remains constant about the shoulder-hand line, fixed γ in fig. 1.

This geometric constraint was adjoined to the inverse kinematics procedure by way of describing the four joint angles when the links are both in the vertical plane, then rotating the plane about the shoulder-hand axis by an angle γ. This was accomplished using Rodrigue's formula about an arbitrary axis.

3. *Small variation in upperarm-forearm plane*. Experimental results indicated the upperarm-forearm plane exhibited a small rotation for a particular task. This led

us to permit γ to vary in accordance with the direction of the velocity vector of the hand. The angular variation of the vector tangent to the path of the hand was recorded from experimental data then scaled and input to the Rodrigue's formula as the variable γ.

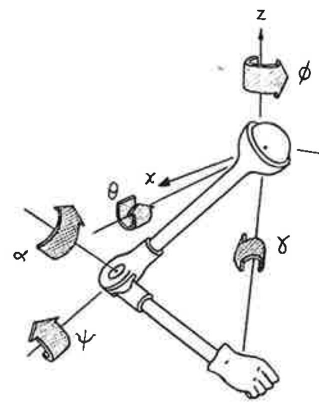
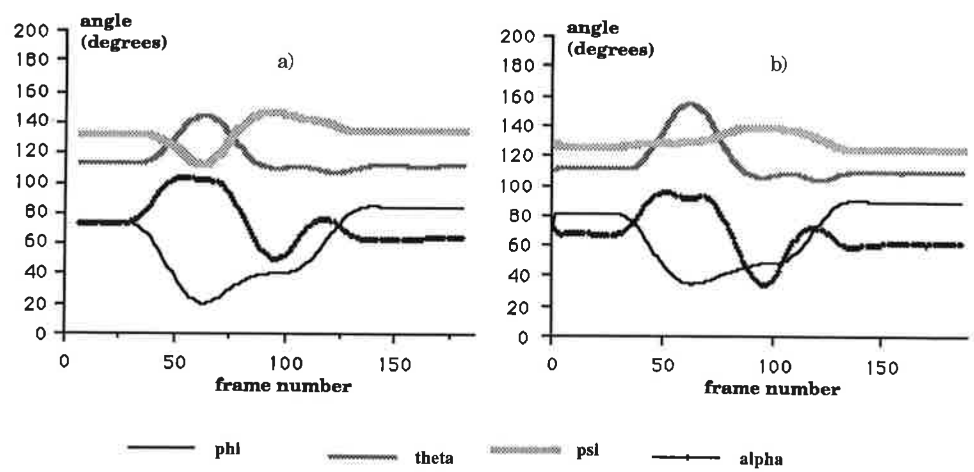


Fig. 1 Configuration of arm with γ- constraint angle

EXPERIMENT

The experimental part involved monitoring the arm motion of a human subject for four routine tasks; arm curl, drinking, swinging, pulling. The system used was the WATSMART system that employs LED's placed on the arm and gives xyz information of the emitters.

The data were collected at 40 Hz, filtered at 4 Hz, then mapped to joint space and differentiated twice using finite difference to yield velocity and acceleration. Fig. 2 shows a comparison between actual data from the experiment and the three constraint methods, for joint angular displacement for the 'swinging' motion which is the most wide ranging.



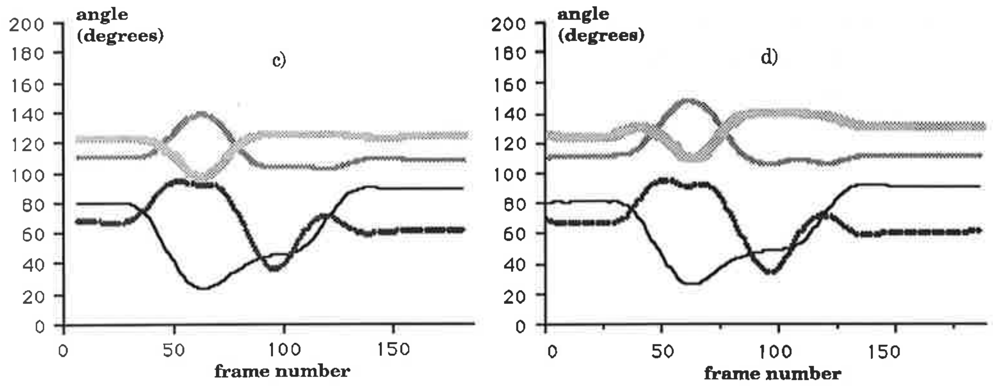


Fig. 2 Angular trajectory for the 4 D.O.F. a) experimental b) pseudoinverse c) fixed γ d) variable γ

RESULTS AND CONCLUSIONS

From fig. 2 it is evident that for routine motions such as 'swinging' (picking up and moving an object around an obstacle) all three constraints yield motion that is relatively close to the actual. On closer inspection, however, the pseudoinverse constraint veers away from actual motion, particularly in humeral rotation. In the other tasks, pseudoinverse produces motion that is not aesthetic; which leads one to conclude that minimizing travel of joints is not a physiological criterion in itself.

Maintaining γ constant or permitting a small variation, yield motion that is aesthetic, with little variation between the two. This relatively simple constraint in joint space could prove to be a significant factor in resolving arm motion, particularly with a view towards its ease of implementation in a control strategy.

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Electromyographical Kinesiology
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UPPER EXTREMITY FORCE REQUIREMENTS IN VIOLIN VIBRATO: A DYNAMIC ELECTROMYOGRAPHIC STUDY

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INTRODUCTION

Musicians are a group of professionals who execute highly skilled complex movements that often require speed, utilization of joints at extreme limits and assumption of awkward unnatural postures. For example a pianist may execute as many as 600 distinct motor actions per second within the course of playing a musical piece (1). Violinists execute intricate motions which require well balanced muscle and joint functions. Violin vibrato playing requires well sequenced periodical muscular function involving all the joints of both upper extremities. It specifically requires even, well coordinated "forward and backward movement of the forearm and hand balanced by slight left and right swinging of the elbow caused by minute clockwise and counterclockwise rotation of the upperarm" (2, p25). The latter represents a balanced biomechanical kinematic chain of the whole arm (3). If the chain is balanced, the back and forth swinging of the hand is supported by smooth involuntary slight in and outward rolling movements of the upper arm (4). However, if the chain is imbalanced, either due to joint laxity, poor positioning or inappropriately designed instrument, the arm produces more forceful voluntary push-pull movements. If the latter occurs consistently for a prolonged time overuse syndrome (static work), a risk factor for musculoskeletal injuries and a common one in violinists, may ensue. The imbalance may further lead to tendinitis, carpal tunnel syndrome and or muscle weakness all of which are common problems reported by violinists (5,6,7). Some of these disorders have been serious enough to disrupt violinists performance significantly and/ or permanently.

Although attempts have been made to find solutions to these problems, little is understood about the physiological and biomechanical components of violinists' performance such as the muscle pattern, force requirements and their relationships to musical performance. Therefore, the purpose of this study is to determine the upper extremity musculature force requirements during the performance of violin vibrato. Establishing the normative musculoskeletal parameters of violin vibrato will allow accurate assessment, diagnosis and rehabilitation of injured violinists.

MATERIALS AND METHODS

Seven healthy professional violinists (6 females, 1 male), age 22-50 (M = 39 $\pm$ 17 years; experience = 10 or more years) with no history of neurological, orthopedic or sensory problems participated in the study. Subjects were required to perform 8 vibrato tasks in random order, each lasting 3 seconds. The tasks were performed in two; positions (one close to the frog, and one close to the bridge), four in each, using the middle finger. In each position, all 4 strings: A, D, E, and G; were played. Surface EMG activity was recorded from triceps, biceps, flexor digitorum, flexor carpi ulnaris, pronator teres, flexor carpi radialis, extensor carpi radialis, extensor carpi ulnaris, and extensor digitorum using a multi-channel analog-to-digital converter, sampling at 1000 Hz. Nine sets of surface electrodes, mounted on miniature differential preamplifiers (Motion Control, Inc., Utah, Gain = 150, bandwidth = 95 to 280 Hz), were placed on muscles monitored. All electrodes were connected to to a common power supply unit and then to an A/D board.

A microphone was placed close to the violin and connected to a Fairlight Voicetracker with an MIDI interface (Fairlight Pty. Ltd., Sydney, Australia). Filters were adjusted to accommodate the musical tasks. The DC output of the instrument was connected to the A/D board in parallel to the EMG channels, thus achieving synchronicity. The input level gate was kept relatively high to minimize noise. A few EMG and sound offset were taken to determine the respective channels offsets and correct them digitally.

RESULTS

The EMG signals were periodic simulating vibrato motions, as found in a previous study (8, Fig.1). In the latter study flexor and extensor digitorum were found to be the most directly correlated to sound energy (Fig. 2) during vibrato playing. Besides , from another study, surface EMG - force calibration curves were available for these two muscles for each subject. These curves were used to convert the dynamic EMG signals into force.

All regressions between amplitude of static EMG and muscular force were linear with high R squared value. Therefore, the curves were used to convert the dynamic EMG signals in vibrato playing into force. Averages were obtained per muscle and per task, and subjected to an analysis of variance. The extensor digitorum produced a mean force of 2.73 $\pm$ 0.75, while the flexor digitorum yielded a mean force of 8.36 $\pm$ 2.807, and the two were significantly different ($F = 13.94$; $p = .025$). There was no significant difference between strings and positions.

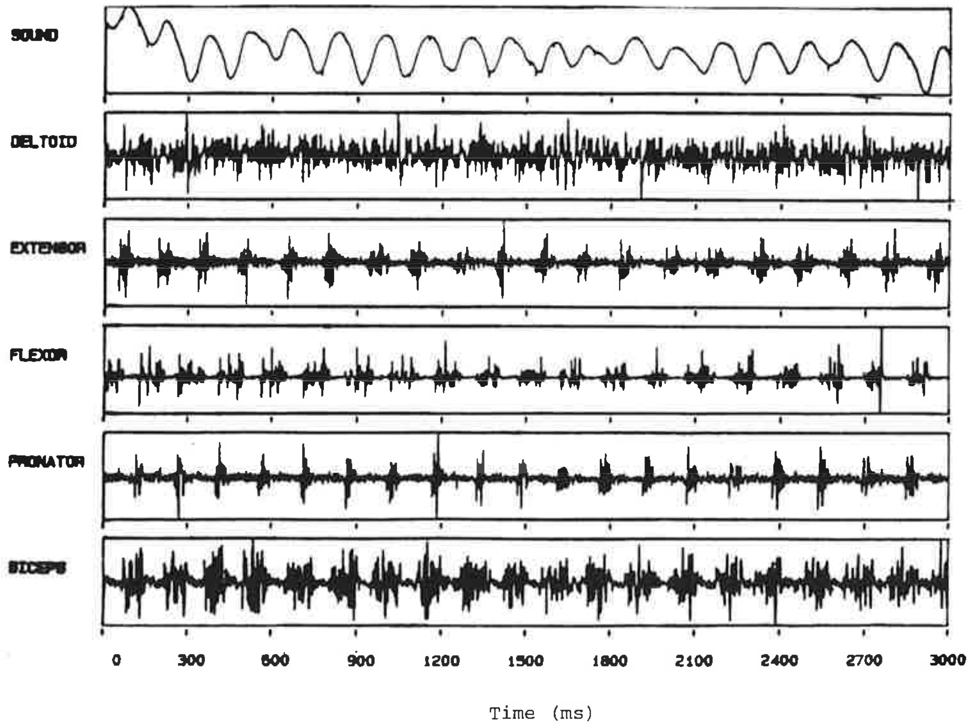

Fig.1. Raw data file showing periodic EMG signals except for deltoid.

CONCLUSION

Objective and reproducible method of quantifying muscle function and force requirements during musical performance may guide early diagnosis and treatment of musical task - related musculosketal injuries. This study validates the use of individual static EMG - force relationship to convert dynamic EMG data into force, as long as the task is periodic and can be analyzed cycle by cycle i.e. violin vibrato. Ongoing analysis is addressing the remaining muscles tested, in terms of force, as well as autocorrelation with the sound signal. Furthermore, the dynamic EMG data is also being correlated with 3D video data collected simultaneously, on the violin and the bow sides.

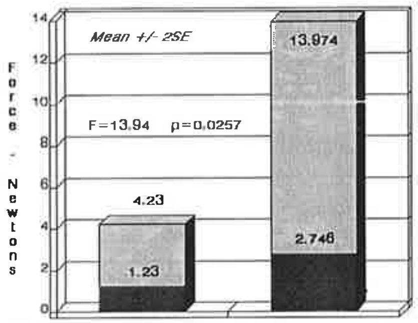

Fig. 2

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

COMPARING DIFFERENTIATING PROCEDURES FOR KINEMATIC DATA

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INTRODUCTION

Accuracy, automaticity and speed represent a design goal for the derivative assessing algorithms. The accuracy of the derivatives is mostly affected by the noise superimposed on the signal. A white gaussian model can be often assumed for the noise and its broad-band nature gives rise to problems related with the high frequency magnification performed by the derivative operators. A low-pass filtering set at a cut-off frequency which reduces high frequency components of noise is thus mandatory. If an estimate of this frequency, for a given data set, exists, a quite fast numerical filtering could be done, but this is usually not the case. Furthermore, the numerical time-domain approximation of the differentiator implies the use of many coefficients and causes wide distortions at the edges of the data record. The problem of the cut-off frequency choice was, in past, solved by iterating trials and errors, but it was a long procedure and the final result was left to the operator. Recently, automatic methods, based on objective criteria, were developed. Two of them will be discussed: the Linear-phase Autoregressive (AR) Model-Based Derivative Assessment (LAMBDA) algorithm (1) and the Generalised Cross-Validatory Splines (GCVS) (2). The choice has been done because these algorithms show far better performances with respect to others (3). The algorithms will be shortly described and some cross-tests will be presented.

LAMBDA

Lambda uses an AR model fitted to the signal in order to get its Power Spectrum Density (PSD) estimate and a set of parameters suitable for a linear prediction of the signal itself. The PSD estimate allows the computation of a cut-off frequency which assures a desired SNR in the filtered signal. The noise power is computed as the average of the Power Spectrum (PS) at high frequencies, where its contribution is supposed to be predominant with respect to the signal. Under the assumption of whiteness, the whole noise PS is thus known and the cut-off frequency can be set at the value for which the signal PS equals P times the noise power. P is the desired

SNR at the cut-off frequency. The linear prediction parameters allow an extension of the signal before and after the data record bounds, reducing the edge effects due to filtering. It must be noted that this operation does not distort the signal PSD because the linear prediction parameter allow to extrapolate data with the same PSD of the signal. After the extension the data are transformed in the frequency domain, were the filtering is performed, via FFT. The filter has a variable impulse response length, equal to the extrapolation performed. The filter length depends on a parameter (SF) which accounts for the ideality of the low-pass frequency response. The ideal low-pass filter is represented in the frequency domain by a rectangle; increasing SF the filter response approaches the rectangle. The time-domain filter has been expressly designed in order to be suitable for derivative assessment (continuously decaying sidelobes and narrow main lobe)(4). The windowed filter impulse response is transformed in frequency domain and multiplied by the signal transform. The multiplication by $j\Omega$ and $-\Omega^2$ gives, after inverse FFT, the first and second d derivatives.

GCVS

The generalised cross-validation criterion (GCVC) was applied to splines functions by (5). Differently from Reinsch smoothing procedure (6-7), the extent of smoothing is controlled by an objective criterion which does not involve operator's choices. GCVC is based on a regularised version of the spline estimation problem. Regularisation concept (8) allows to damp high frequency oscillations in the solution of an ill-posed problem by controlling a smoothing norm of the data (9). GCVC (10) is suitable for the selection of the regularisation parameter for the derivative assessment problem (9). This parameter was chosen by searching the smoothing function minimising the root mean square error of the prediction of a point, while fitting all the data except that point. An extension to B-splines of the GCVC, GCVS, was presented by (2). The main problems related to this approach are time consumption, the need of long data records ($N \geq 40$) and the naturality condition for the splines. This assumes that for a (2m-1)th order spline, the derivatives of order greater than m-1 must be nil at the signal edges. This is not always true leading to distortions which, not only affect the data record edges, but propagate also inside it. Five order GCVS were chosen thus derivatives of order greater than 2 must be zero at edges.

TEST DATA

Data taken from literature, acquired in laboratory by means of the ELITE system for motion analysis and synthetic functions have been used for the tests. The literature data are the classical wrist angular displacement reported in (11) and (12) and the free falling ball vertical displacement proposed by (13). Among the data acquired with the ELITE system, the vertical displacement of a thrown free falling ball have been chosen. The synthetic data are sinusoids affected by noise and the third order polynomial proposed by (14).

RESULTS

As a paradigmatic result, figure 1 reports the results obtained with GCVS (5th order) and LAMBDA, on the vertical displacement of a launched falling ball sampled at 50 Hz. Figure 2 is relative to the result of the same procedures as above applied to the synthetic function of (14). The results point out a better performance

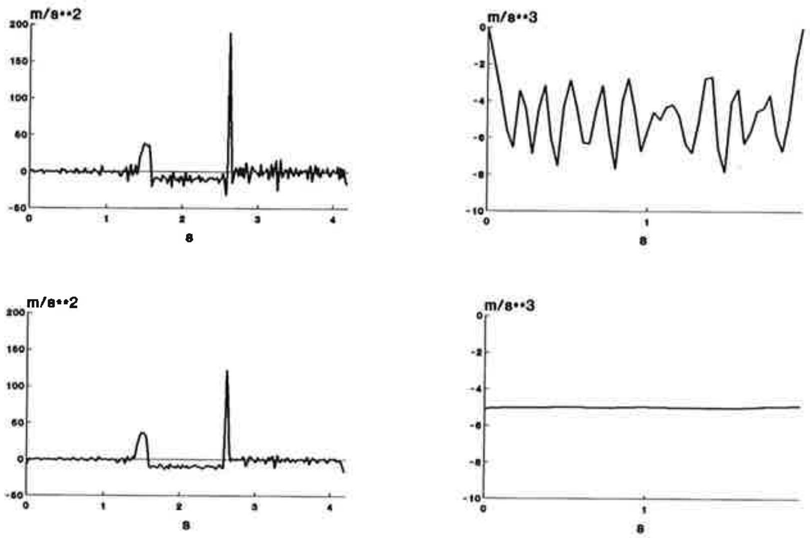

Fig. 1. Falling ball data, second derivative   Fig. 2. Synthetic data, third derivative
(upper panels GCVS, lower panels LAMBDA)

(dramatic in figure 2) of LAMBDA in noise rejection, while GCVS tends to privilege the acceleration peak amplitude (however only 25% more than LAMBDA). A similar behaviour was found in all the other data considered. LAMBDA shows a better or similar behaviour in the second derivatives and a very good agreement of the residual with known or estimated noise. The time consumption for the processing is far lower for LAMBDA. The data reported in the figures took 220 s against 4.8 s (figure 1) and 27 s against 2.5 s (figure 2) for the processing with GCVS and LAMBDA respectively.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

THE VALUE OF CONTEXT VARIATION FOR APPLIED MOTION ANALYSIS

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INTRODUCTION

The present text is primarily focused at the clinical assessment of motor dysfunctions, particularly in rehabilitation practice. Because clinically valid motor assessments require a theoretical framework which also includes non-motor aspects of human motor skills (1), this paper is based on a theory of human performance, rather than on traditional kinesiology.

By discussing the use of simple versus complex environmental conditions in the assessment of motor dysfunctions it is shown that, when simple conditions reveal the absolute level of motor output control, complex conditions give information about the automaticity and neural flexibility of motor control.

In the literature on human skills motor control is regarded as the result of a fine-tuned interaction between perceptual, cognitive and motor processes (2-3). Before, during and after the execution of a movement, several stages of information processing are active to prepare, initiate and evaluate the intended movement. The extent and speed of information processing is strongly related to the complexity and novelty of the movement.

Although in healthy individuals an information processing approach may be more useful for explaining a complex grasping movement than for explaining a highly automated skill, such as standing, in case of motor dysfunctions, formerly automated motions may have become new and complex again. It is assumed that such a breakdown of a skill induces the need of extra perceptual information and attentional resources. Hence, manipulating the perceptual and cognitive variables by creating complex conditions is a necessity for the assessment of automaticity and flexibility of motor control.

DUAL TASK INTERFERENCE IN BALANCE ASSESSMENT

The use of an information processing approach to the assessment of motor dysfunctions will be illustrated by regarding the automaticity of balance control in two types of pathological conditions. The need of attentional resources in order to maintain a quiet upright posture was estimated by the interference of a concurrent non-motor task with a primary balance task, thus creating a dual task procedure.

Subjects were instructed to stand as still and symmetrically as possible for a period of 30 seconds on a dual plate force platform, which recorded the vertical ground reaction forces at a sampling rate of 20 Hz. After a 15-s-period the concurrent task was introduced: 25 coloured words representing colour names were projected onto a screen 1.5 meter in front of the subject (Stroop task). However, the colour names were always incongruent with the printed colours. Subjects were instructed to name the colours of the printed inks as quickly as possible during the remaining 15 seconds, while emphasis was given to maintain the same balance strategy. At the same time they had to suppress the strong tendency to read the colour names. Before every balance test procedure the Stroop test was first practised and then recorded in a sitting position as a single task performance.

By means of a microprocessor the displacements of the centre of pressure (CP) were calculated, filtered (cut-off frequency 5 Hz) and differentiated. The root mean square of the CP-velocities served as a measure of the efficiency of balance control before and after the introduction of the concurrent task. The number of items scored on the Stroop task served as a measure of the concurrent task performance.

Figure 1 shows an individual 30-s-registration of one of the patients. The introduction of the concurrent Stroop task is indicated at the 15th second. It is shown that the last 10 seconds were taken to

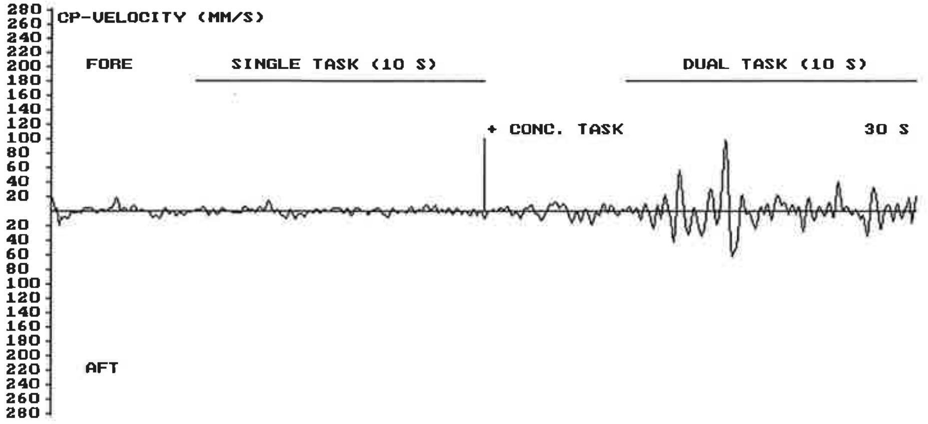

Fig. 1. A 30-s-balance registration (fore-aft sway) of a HMSN-patient wearing new rehabilitation footwear. Note the difference between the single and dual task performance.

represent balancing in a dual task condition. This 10-s-period was compared to the 10 seconds before the introduction of the Stroop task, which represented balancing in a single task condition.

CLINICAL EVIDENCE

The automaticity of balance control was studied in two different groups of patients with peripheral neuromuscular lesions. None of the patients suffered from a marked perceptual or cognitive deficiency.

Lower limb amputees were tested because they suffered from an acute balance disorder, forcing the central nervous system to a fast sensory-motor reorganization. Therefore, a breakdown of automaticity of balance control was expected, which should be less severe after a successful rehabilitation process.

In addition, patients suffering from hereditary motor and sensory neuropathies (HMSN) were tested because they had a slowly progressive disease. Due to a predominantly distal paresis of the leg muscles, sensory impairments and deformities of the feet and ankle joints (pes cavus and equinovarus), these patients suffered from a chronic balance disorder. Because of the foot deformities and ankle instability, most HMSN-patients were provided with individually designed footwear (in its try-out phase called rehabilitation footwear), which consisted of a corrective insole and, in case of varus deformity, of an ankle stabilizing socket together with a roll-off correction under the shoe.

Because of the slowly progressive character of HMSN, the central nervous system should be able to adjust on-line to the peripheral impairments. Therefore, relatively automated balance control was expected in case of barefoot standing. However, when a patient had to adapt to new rehabilitation shoes, a fast reorganization became necessary more or less comparable with the situation in amputees. In this case, a (temporary) loss of automaticity of balance control was expected.

Indeed, in comparison with healthy subjects (matched for age and sex), lower limb amputees showed significantly more dual task interference just after the first training with the definitive type of prosthesis. This interference occurred in both directions of sway, which is in agreement with the fact that a lower limb amputation affects postural stability in the sagittal plane as well as in the frontal plane. Although the degree of dual task interference decreased during the rehabilitation process, this diminution was not entirely significant. However, just before the completion of the rehabilitation process the dual task interference of the amputees on the fore-aft sway was no longer significantly different from the control group.

Barefooted HMSN-patients showed no significant difference of dual task interference in comparison with their matched controls. Conversely, when standing with new rehabilitation footwear, they showed a marked interference in comparison with themselves standing barefoot. This greater interference was significant only with respect to the fore-aft sway, which is congruent with the notion that ankle stabilizing shoes mainly affect postural stability in the sagittal plane.

DISCUSSION

This paper is not meant to lay specific emphasis on balance control in amputees or HMSN-patients. These groups of patients serve to illustrate the phenomenon of dual task interference in clinical motor assessment and the use of an information processing approach to applied motion analysis in general.

It has been shown that interference of an attention-demanding task with balance control can well be predicted from a process-oriented viewpoint. The degree of novelty of the task requirements indicates the observed dual task interference, which implies that this phenomenon is not disorder-specific. Consequently, the interference effects can be interpreted as a lack of automaticity of balance control in a specific situation.

Particularly in rehabilitation medicine, it seems important to be able to estimate the degree of automaticity besides the absolute level of eg. locomotor skills. In the first place, the degree of task automaticity is indicative of the stage of a learning process. In the second place, it is theoretically related to the safety of motor performance in daily life, which is always a succession of more or less complex situations. It should be realized, however, that this type of information can only be obtained by a standardized way of context variation in applied motion analysis.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

MOTOR UNIT DISCHARGE BEHAVIOR OF HUMAN TIBIALIS ANTERIOR MUSCLE AT THREE ANKLE POSITIONS

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INTRODUCTION

Purposeful movement in humans requires that muscle activation be controlled under a variety of external conditions that effect the contractile properties of muscle fibers. Contractile properties of muscle are affected by the contractile history of the muscle, fatigue, and muscle length. Adjustments in motor unit discharge characteristics have been shown to occur during fatigue to accommodate for slowing of the contractile process.[1] The central nervous system may make similar adjustments to accommodate for contractile changes imposed by shortened and lengthened muscle. Although recruitment threshold of human masseter motor units has been shown to increase with increases in passive tension as muscle is lengthened[2], little information is available on how muscle length may influence motor unit discharge behavior during voluntary contractions. The purpose of this investigation was to evaluate the relationship between muscle length and motor unit behavior in human tibialis anterior muscle. The results revealed that motor unit discharge behaviors during isometric contractions up to 40% MVC were adjusted in association with changes in ankle joint angle and muscle length.

METHODS

Twenty untrained male subjects (ages 21-35) with no history of injury or orthopaedic abnormality of the lower extremities participated in this study. Tibialis anterior motor unit action potentials were recorded by fine wire electrodes during ramp up-steady hold-ramp down isometric contractions at 30 degrees of ankle plantar flexion (lengthened), midway between plantar flexion and dorsiflexion (neutral), and at 20 degrees of ankle dorsiflexion (shortened). Subjects increased dorsiflexion torque at 5% MVC/sec to steady torque of 10%, 20%, 30% or 40% MVC for eight seconds, then decreased torque at 5% MVC to baseline. Dorsiflexion torque and motor unit action potentials were recorded on FM tape and analyzed off-line. Units were discriminated with a window discriminator and interspike intervals (ISIs) were calculated using a microcomputer.

RESULTS

One-hundred thirty-six units from 16 of the 20 subjects in the study were analyzed. As it was not possible to follow a single unit across different ankle positions and contraction levels, all units recorded were considered unique and entered into pooled data for statistical analyses. Because the waveform of motor units sometimes changed as torque was increasing or decreasing, not all units could be reliably discriminated throughout the task. As a result some portion of the total number of units was used for analyses at recruitment, steady-state torque, and derecruitment.

A representative plot of interspike intervals and dorsiflexion torque is shown in Figure 1. In this 40% MVC trial, the unit was recruited at 3.06 N.m of torque with an initial ISI of 129 ms. A mean interspike interval of 55.8 ms was found during steady-state torque of 7.48 N.m, and the unit was derecruited at 5.23 N.m of torque with a derecruitment ISI of 244 ms.

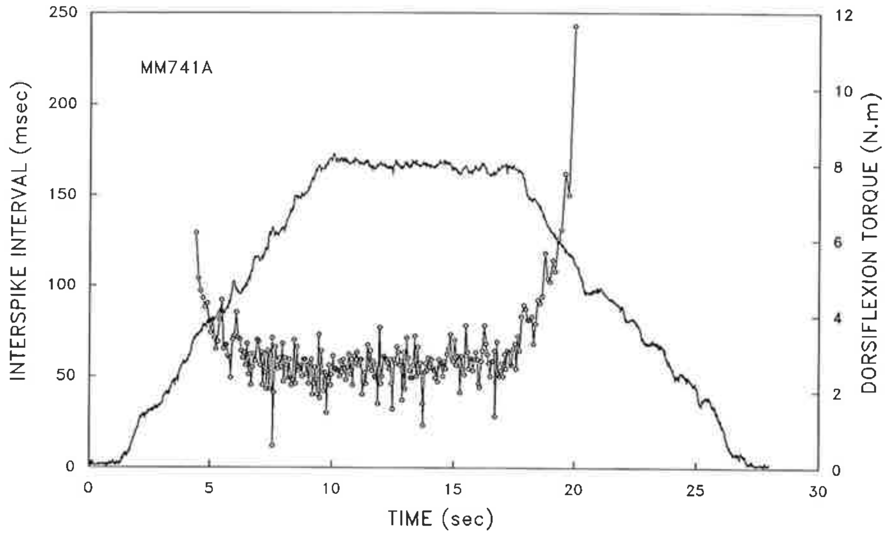

Figure 1. Representative plot of ISIs and dorsiflexion torque for a 40% MVC contraction.

One-hundred sixteen of the 136 units were used to evaluate the effect of ankle position on motor unit discharge behavior at and immediately after recruitment. Mean instantaneous discharge rate at recruitment (reciprocal of the first ISI) was 30.6 pps, 16.4 pps, and 13.0 pps when the ankle position was 20 degrees dorsiflexed (DF), neutral, and 30 degrees plantar flexed (PF), respectively (Figure 2). An ANOVA revealed that the mean recruitment discharge rate was significantly greater at 20 degrees DF than at 30 degrees PF ($p < 0.05$). Further analysis of the data revealed that in some spike trains, the initial interspike interval was brief, with a duration of 40 ms or less. Brief recruitment ISIs were observed in 10 of 41 units at 20 degrees DF, in 3 of 34 units at neutral, and in 3 of 43 units at 30 degrees PF. A Chi-square analysis revealed that differences in the frequency of occurrence of brief initial interspike intervals were significant across muscle lengths ($p < 0.05$).

Data from all 136 units were used to determine the effect of ankle position on motor unit discharge rate during the steady torque portion of the trial. During steady torque conditions, the relationship between mean discharge rate and torque was described by a linear regression for each ankle angle. The slopes of the regression lines were 0.67 pps per N.m, 0.39 pps per N.m, and 0.27 pps per N.m for 20 degrees DF, neutral, and 30 degrees PF, respectively. The slopes of the linear regressions describing the relationship between discharge rate and torque were significantly different ($p < 0.05$), with the slope significantly greater for the 20 degree DF position than for either the neutral or the 30 PF position.

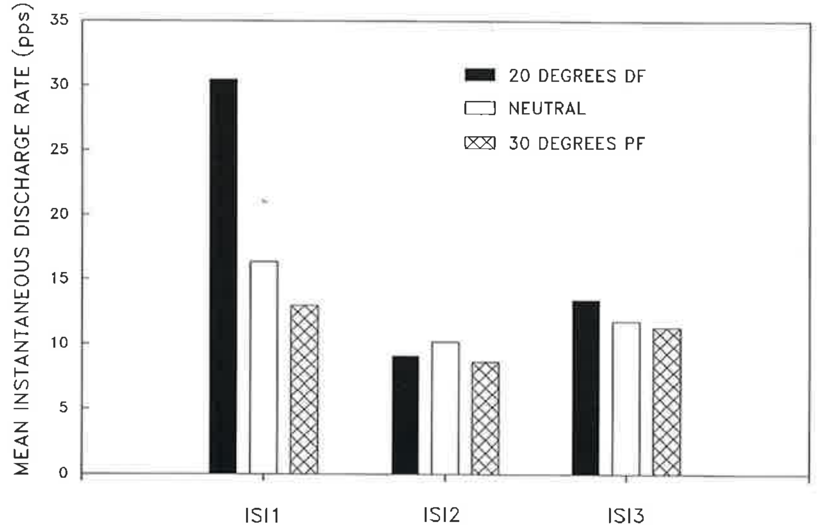

Figure 2. Mean discharge rate for ISI1–ISI3 for three ankle positions.

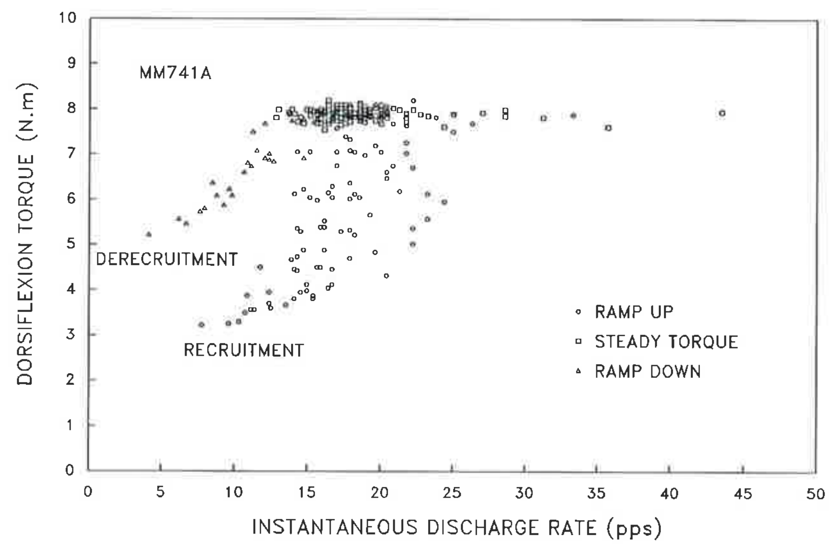

Figure 3. Plot of discharge rate hysteresis during increasing and decreasing isometric torque.

Ninety-five of the 136 units were used to evaluate discharge behavior when units ceased firing. Mean derecruitment discharge rate was 5.3 pps, 6.4 pps, and 4.8 pps for 20 degrees DF, neutral and 30 degrees PF, respectively. For all ankle positions, units ceased discharging at a lower rate than at recruitment. In addition, a hysteresis of discharge rate was observed during increasing and decreasing torque conditions, with the discharge rate being less during decreasing torque than during the increasing torque portion of the trial (see Figure 3 as an example).

DISCUSSION

Ankle position was shown to influence the initial discharge frequency of motor units of the tibialis anterior muscle during isometric ramp contractions. This finding is evidence that the central nervous system may adjust the output of the alpha motor neuron pool to discharge more rapidly at recruitment when the muscle is shortened (20 degrees DF). This adjustment may help take up the slack in passive elements of resting fibers when the muscle first begins to contract.[3]

During the steady torque portion of the trial, motor units discharged more rapidly to produce each N.m of torque when the ankle was at 20 degrees DF (muscle shortened) than at neutral or 30 degrees PF (muscle lengthened). This may be evidence that the tibialis anterior muscle is operating on the steep portion of the angle-torque curve when at 20 degrees DF, where a higher discharge rate is necessary to produce the same amount of torque than on the plateau of the angle-torque curve.[4,5]

The discharge rate hysteresis observed at all muscle lengths is evidence that torque was maintained with a lower discharge rate as torque was decreasing than when torque was increasing (Figure 3). A similar hysteresis has been reported in cat soleus and medial gastrocnemius muscle using a linearly increasing and decreasing stimulation frequency paradigm.[6] The higher discharge rate observed during increasing torque may take advantage of the catch-like properties of muscle[7], whereby torque can then be maintained with a lower discharge rate during decreasing torque conditions.

ACKNOWLEDGEMENTS

This work was supported in part by the Foundation for Physical Therapy and by Dr. Kukulka's NIH Grant NS24991.

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Electromyographical Kinesiology
P.A. Anderson, D.J. Hobart and J.V. Danoff, editors.

KINETIC LIFTING REQUIREMENTS OF HOSPITAL WORKERS USING ON SITE 3D VIDEO ANALYSIS

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INTRODUCTION

Low back injury is a common problem that affects more than half of the working population at some time during their working career (1,2,3). Approximately 23.8% of all high cost workman's compensation accidents have been associated with manual materials handling. Epidemiological studies indicate occupational factors such as lifting as a major contributor to low back injury, yet its cause remains obscure (2,4). Furthermore, lifting assessment has been focused on establishing safe lifting limits for the average person. However, safe and accurate lifting assessment need to be personalized according to the anthropometry and the physical characteristics of each lifting task (4,5). Lifting methods have frequently been taught in back programs to reduce the incidence of low back injury. However, assessment of lifting capacity have been limited to industries and laboratory settings. In order to design a job to match the physical capabilities of the majority of workers and to place a worker on a job that does not exceed his/her lifting capacity job site lifting requirements need to be established. The purpose of this study is twofold: to determine the kinetic requirements of various lifting tasks in a hospital setting, and to introduce a model for assessing three tasks in the workplace.

REVIEW AND THEORY

Surface electromyography has been widely used to study job requirements (6). Knowledge of the biomechanical profiles of lifting - isometric, isokinetic and isoinertial - have evolved (7,8). Jonsson (7) popularized the use of electromyographic kinesiology and quantified EMG in musculoskeletal performance analysis and ergonomic workplace. Almstrom & Kadefors (8) developed a portable equipment for the measurement of EMG during a normal work situation, consisting of lifting and lowering tasks. Because

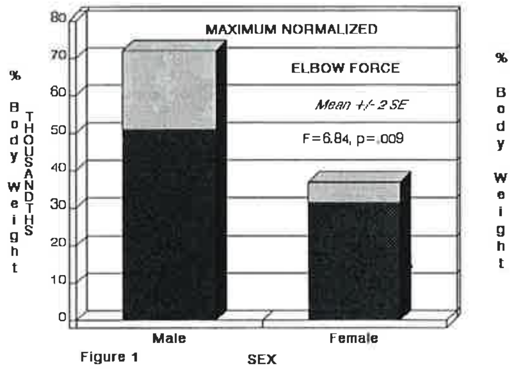

Figure 1

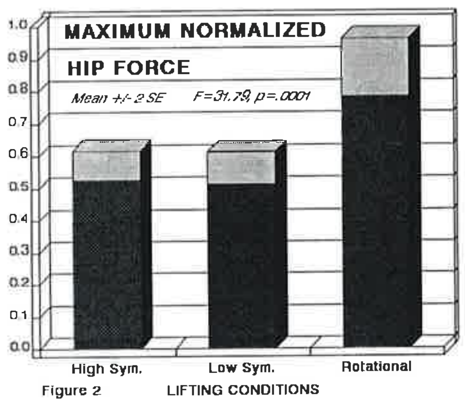

Figure 2

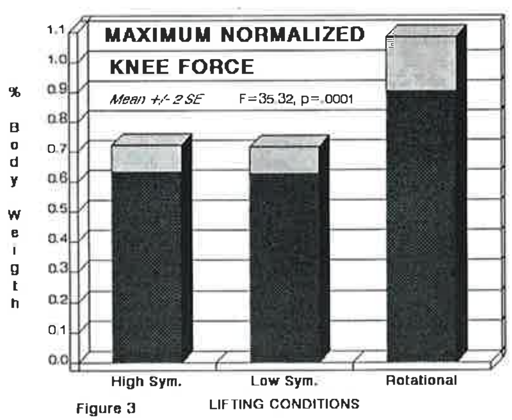

Figure 3

most muscles move more than one joint, and because most activities involve multiple and complex joint motions in more than one plane, a biomechanical profile of a given task cannot be complete without video analysis, preferably 3-D, in the workplace. Hence the need for the present study.

METHODS

Twelve adults (7 males, 5 females) aged 20-45 (M = 44, SD = .13) participated in the study. Subjects' personal and job related data were acquired via a standard occupational questionnaire . The subjects were asked to randomly freely perform the following lifting tasks, with loads weighing 10% and 20% of their body weight, while being videotaped by 3 cameras in 3 different planes, after on-site calibration:

1. lift symmetrically from the ground to a table height of 52 cm (low Sym.).
2. Lift symmetrically from the ground to a table height of 95cm (high Sym.).
3. lift from a table height of 95cm to a standard shopping cart, 35cm deep, located to the left of the subject, perpendicularly to the table (Rotational).
4. Repeat 1, 2, and 3 ; both from the ground- to- table or cart to table (Lift Up) and the table-to-ground (cart) (lift down) directions. Kinematic data were recorded by the video cameras. A special software was used to digitize the video data and to obtain a 3-D reconstruction (Ariel Dynamics Inc., 1988). The kinematic data were obtained for the respective joints and for each task. Kinetic data were derived from the kinematic data using an estimated ground reaction force and the subjects' anthropometric data. The kinetic data were subjected to a multiple analysis of variance.

RESULTS

Figures 1 to 3 depict the findings in this study. Maximum shoulder range of motion was significantly different in all the lifting conditions. Maximum velocity of the shoulder and hip were significantly different both in all lifting conditions and lifting directions. Significant difference existed in maximum shoulder and hip acceleration for all lifting conditions and lifting directions. Knee and ankle acceleration were significantly different only in the lifting directions.

Maximum shoulder force was not significantly different across tasks (M = 8.25 ± 0.23% of body weight). Maximum elbow force was significantly different between sexes (see fig.1). Maximum hip and knee force were significantly different in both lifting directions and lifting conditions. However, the difference in the latter was stronger (see figs 2 & 3). The trunk force is considered to be about twice the hip force in this model.

DISCUSSION

Lifting conditions, especially lifting directions, demonstrated the most significant differences between the kinematic variables,primarily their maximum values. Findings indicate that the kinetic requirements of the shoulder and the hip show the greatest difference as compared to the other joints. This might be due to the characteristics of the lifting tasks. It is noteworthy that the subjects spontaneously did not bend their knees in all the tasks. Velocities were significantly higher at the knee and the hip during lift up, and lower for the shoulder. Maximum accelerations did not exceed 0.4 to 0.5 g (gravity) at all joints.

CONCLUSION

This study illustrates a model for determining kinematic and kinetic lifting requirements, in the workplace itself, using portable video cameras. Furthermore, kinetics were derived from the kinematics via software. The introduction of portable 3-D videoanalysis in the workplace have numerous applications in the fields of ergonomics, sports, and occupational and rehabilitation medicine. It can have direct therapeutic implementation. Information obtained from the 3-D can be analyzed and programmed into a software of computerized exercise machine that can be used for specific training of workers.

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